Kinesthetic Force Feedback and Belt Control for the Treadport Locomotion Interface

Babak Hejrati, *Student Member, IEEE*, Kyle L. Crandall, John M. Hollerbach, *Fellow, IEEE*, and Jake J. Abbott, *Member, IEEE*

Abstract—This paper describes an improved control system for the Treadport immersive locomotion interface, with results that generalize to any treadmill that utilizes an actuated tether to enable self-selected walking speed. A new belt controller is implemented to regulate the user's position; when combined with the user's own volition, this controller also enables the user to naturally self-select their walking speed as they would when walking over ground. A new kinesthetic-force-feedback controller is designed for the tether that applies forces to the user's torso. This new controller is derived based on maintaining the user's sense of balance during belt acceleration, rather than by rendering an inertial force as was done in our prior work. Based on the results of a human-subjects study, the improvements in both controllers significantly contribute to an improved perception of realistic walking on the Treadport. The improved control system uses intuitive dynamic-system and anatomical parameters and requires no *ad hoc* gain tuning. The control system simply requires three measurements to be made for a given user: the user's mass, the user's height, and the height of the tether attachment point on the user's torso.

Index Terms—Immersive environment, haptic interface, locomotion interface, treadmill, tether, control

1 Introduction

Treadmills are commonly used in locomotion interfaces to enable users to walk through virtual environments, and are widely used in physical exercise and gait rehabilitation [1], [2], [3]. Although treadmill speed has traditionally been set by manual control, the trend is for self-selected speed adaptation by measurement of user position or some other form of user intent. This provides for the belt speed to be instantaneously set by a user, leading to a more natural locomotion experience. The task for the treadmill controller is to achieve accurate and stable belt motion, whether the user is walking or running, going forward or backward, or starting or stopping.

We have been developing one particular locomotion interface, the Treadport (Fig. 1), whose key features include the following: (1) a large belt (1.8×3 meters); (2) a six-axis mechanical tether attached to the back of a user wearing a harness, which is used to measure body position and orientation, to control belt speed, and to apply horizontal kinesthetic force feedback to the user [4]; and (3) a six-degree-of-freedom mechanism-based harness with a telescoping spine to accommodate the complex motion of the user's back without slipping, and with the ability to change the point of force application of the mechanical tether to the user [5]. There is a safety dead-man switch held by the user throughout locomotion on the Treadport, and if the user wants to stop the system for any reason, it can be done by simply releasing the switch. The controller of the Treadport is implemented in dSPACE1103. Other characteristics of the Treadport that are not utilized in the present study include: programmable vertical weight support; a CAVE-like [6] visual display; the ability to turn in the virtual environment [7]; and a wind display system [8].

A key difference between the Treadport and most other treadmill-style locomotion interfaces is the presence of the mechanical tether, which can generate forces on a user via a harness [9]. The kinematics of the mechanical tether (see Fig. 1) include a two-axis rotary joint at the base (sensed with potentiometers, but not actuated), a prismatic joint (sensed with an optical encoder, and actuated with a brushed DC motor), and a three-axis rotary joint at the attachment point with the user's harness (sensed with potentiometers, but not actuated). Without a mechanical tether, it is not possible for treadmill locomotion to be energetically realistic when the user's body remains nearly stationary with respect to the ground, since inertial forces due to body acceleration are missing. Previously, Christensen et al. presented a tether controller that implemented direct inertial force feedback [9]. With recent application of the Treadport to rehabilitation of patients with spinal-cord injury [10], limitations of this controller became apparent due to the fragile walking conditions of these patients. It was felt too difficult to start walking by pulling against the tether, stopping was sometimes unnatural due to the improper recentering controller, and the apparent inertia felt too large.

To address the issues described above, in this paper we present a new kinesthetic-force-feedback controller based on maintaining the user's sense of balance as the belt moves under their feet, regardless of the underlying belt control

- B. Hejrati, K.L. Crandall, and J.J. Abbott are with the Department of Mechanical Engineering, University of Utah, Salt Lake City, UT 84112, USA. E-mail: {babak.hejrati, kyle.crandall, jake.abbott}@utah.edu.
- J.M. Hollerbach is with the School of Computing, University of Utah, Salt Lake City, UT 84112, USA. E-mail: jmh@cs.utah.edu.

Manuscript received 22 July 2014; revised 10 Feb. 2015; accepted 11 Feb. 2015. Date of publication 15 Feb. 2015; date of current version 15 June 2015.
Recommended for acceptance by R. Adams.
For information on obtaining reprints of this article, please send e-mail to: reprints@ieee.org, and reference the Digital Object Identifier below.
Digital Object Identifier no. 10.1109/TOH.2015.2404357

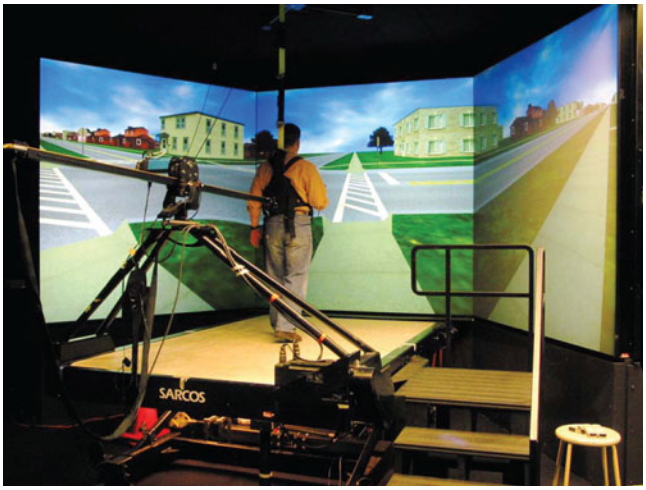


Fig. 1. Treadport locomotion interface. For clarity, the system is shown before the addition of the wind-display system.

system. We revisit the Treadport's belt controller by implementing a previously published controller from a non-tether treadmill system and determining the dynamic properties of that controller that result in the best sense of realism within the Treadport. We also present an adaptive dead-zone algorithm that enables users to stand still on the belt when desired without fear of the belt moving unintentionally. A stability analysis is conducted to ensure the stability of closed-loop system with a user with their own volition in the loop. In human-subject experiments, our new Treadport control system achieves a sense of stable and smooth acceleration and deceleration when moving forward or backward, stopping slowly or quickly, and when standing (and swaying) comfortably without causing involuntary motion of the belt. We demonstrate with statistical significance how users' perception of realistic walking is enhanced by using our new controller: the kinesthetic-force-feedback controller is compared to the previous controller, and the most-preferred dynamic properties of the belt-speed controller are determined in a separate experiment. Although experiments are conducted with the unique Treadport locomotion interface, the results in this paper will generalize to any locomotion device with self-selected speed and a tether capable of applying horizontal forces to the user.

Several methods have been used previously to estimate and generate a user's self-selected walking speed on a treadmill belt. In [9], the desired velocity of the treadmill belt was derived from a proportional-integral (PI) controller. A similar PID approach is taken in [11]. The PI and PID gains were chosen through an *ad hoc* procedure and remained constant for all users. More recently, [12] proposed a second-order dynamic observer to estimate the desired belt velocity. A different approach is taken in [13], in which a rigid bar with a force sensor is attached to the user, and user force against the bar is used to generate belt speed through an admittance controller. Gait parameters such as ground reaction forces [14], [15], [16] and foot-swing velocity [17] can be used to update the treadmill speed.

The major challenge for treadmill-style locomotion interfaces for simulating overground walking is dealing with the acceleration of the belt, which affects a user's stability since it exerts forces on the user that would not be felt during overground walking. Recently, Souman et al. [12] used a position controller combined with a dynamic observer for estimating voluntary walking speed on a 6-meter-long treadmill. A goal of their research is to present realistic vestibular stimulation during acceleration, and consequently their treadmill belt is long in order to allow real acceleration before reaching the front of the belt. If their controller is applied to smaller treadmills, a user would feel undesired large inertial forces [18]. Most recently, Kim et al. [18] proposed an estimation limiter to attenuate the unwanted inertial forces due to acceleration/deceleration of the treadmill belt. Of course, it is not possible to completely eliminate these unwanted forces with any belt of finite length.

Energy expenditure on treadmill devices is another important consideration. Frishberg et al. [19] showed that sprinting on a treadmill requires significantly less energy than sprinting on the ground. Lee et al. [20] found that global patterns including kinematics and kinetics are similar between overground and treadmill locomotion, however the energy cost, regardless of the method used to compute it, is significantly different. Crétula et al. [21] showed that for computing mechanical work during treadmill locomotion and comparing it with overground, variations in belt speed must be taken into account. Acceleration/deceleration of a treadmill belt due to speed adaptation requires less energy expenditure, which can be compensated using kinesthetic force feedback.

2 Control Algorithm

The objectives of the Treadport's controller design are to achieve the user's intended self-selected walking speed while mitigating the obtrusive and unnatural effects of belt acceleration on the user, to create a natural walking experience similar to overground locomotion. To fulfill these objectives, two separate controllers have been implemented to work together with the human user as a complete closed-loop system: (1) A recentering controller regulates the user's position to some reference position on the belt (typically near the center), which ultimately provides an instantaneous desired belt velocity command to a low-level belt-speed controller (the low-level belt-speed controller, as well as an adaptive dead zone to improve system behavior when the user is attempting to stand still, are included as supplemental material). (2) A kinesthetic force-feedback controller exerts a horizontal force on the user's torso via a mechanical tether in order to create a stable and energetically realistic walking experience. In this section, we describe our recentering controller and our kinesthetic force-feedback controller. We then analyze the stability of the combined system, with a human user in the loop.

2.1 Recentering Controller

The principle of self-selected speed, in its most basic form, is quite simple: a user walking on the belt should be kept near some reference position (typically near the center of the belt), and if the user advances beyond the reference position it indicates that the user's intent is to increase walking speed, so the belt speed is increased until the user is brought back to the reference position with a new equilibrium

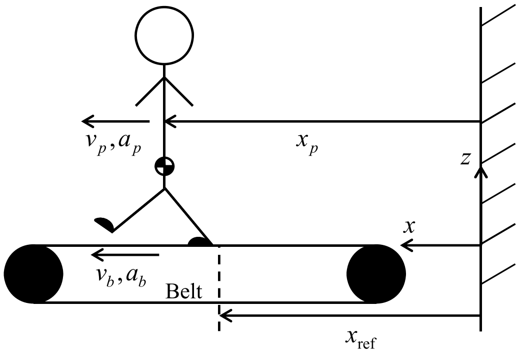


Fig. 2. A user on the Treadport. The user's position x_p at the tether attachment point and the reference position x_{ref} are measured with respect to the same arbitrary inertial reference frame. The belt's velocity v_b and acceleration a_b are defined as positive in the forward direction.

walking speed, with deceleration handled analogously. Any scheme that accomplishes this goal could be a valid self-selected speed controller. It is also necessary on any treadmill belt of finite length that some controller exists to prohibit the user from walking off the edge of the belt. In practice, a well-designed recentering controller also serves as a self-selected speed controller once a human user, with the ability to establish their own self-selected gait pattern, is included in the closed-loop system.

Because human users have their own volition, a user may choose to stop walking by planting their feet instantaneously at any moment, and in that moment the recentering controller must act to bring the now-riding user to a stop quickly and safely. This is an additional factor that should be considered when designing the recentering controller.

We implement a simple PD controller for recentering that uses the position error between the user and the reference position to set the desired belt acceleration:

$$a_{bd} = K_x(x_{\text{ref}} - x_p) - K_v v_p, \tag{1}$$

where x_p is the person's position, v_p is the person's velocity, and x_{ref} is the reference position, all in the inertial frame (Fig. 2); a_{bd} is the desired belt acceleration; K_x and K_v are proportional and derivative gains, respectively. Note that x_{ref} is static, so its derivative is always zero. We see that the belt tends to be accelerated backward from the user's point of view whenever they are either in front of the reference position (i.e., too close to the front of the belt) or moving forward in the inertial reference frame (i.e., getting closer to the front of the belt), with an analogous and opposite behavior when accelerating the belt forward.

In the Treadport, the position x_p and velocity v_p are measured by a mechanical tether attached to the user, via the system's forward kinematics. The resulting desired acceleration a_{bd} is then numerically integrated to derive the instantaneous desired belt velocity v_{bd}, which is then given to the low-level belt-speed controller. Equation (1) is mathematically similar to the "second-order controller" in [12]. The belt velocity v_b can be expressed in terms of the user's speed relative to the belt, $v_{p/b}$, and their speed relative to the inertial frame v_p:

$$v_{p/b} = v_p - v_b \quad \leftrightarrow \quad v_b = v_p - v_{p/b}. \tag{2}$$

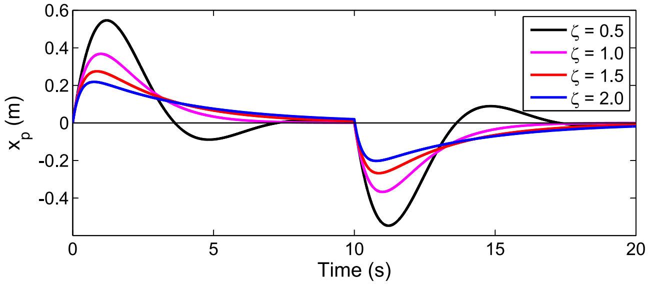


Fig. 3. The effect of different values of ζ while $\omega_n = 1$ rad/s on a user's position x_p during walking. Starting from rest, $v_{p/b} = 1$ m/s for the first 10 sec, and then $v_{p/b} = 0$ m/s.

The user perceives $v_{p/b}$ as their instantaneous walking speed. Assuming that the dynamics of the low-level belt-speed controller are of sufficiently high bandwidth relative to the dynamics of the recentering controller (the bandwidth of our belt-speed controller is an order of magnitude faster than the recentering controllers considered), we can drop the "desired" subscript "d" in the desired speed v_{bd} and acceleration a_{bd} for a simplified analysis. Substituting (2) into (1), the Laplace transform yields:

$$s(v_{p/b} - v_p) = K_x(x_p - x_{\text{ref}}) + K_v v_p \tag{3}$$

leading to the transfer function for the user's position:

$$x_p = \left(\frac{K_x}{s^2 + K_v s + K_x}\right) x_{\text{ref}} + \left(\frac{s}{s^2 + K_v s + K_x}\right) v_{p/b}. \tag{4}$$

We observe that the position of the person on the belt is a function of two independent variables: x_{ref} and $v_{p/b}$. We see that with $v_{p/b} = 0$, which occurs when the user plants their feet and rides the belt, the person will eventually be brought to x_{ref} with no steady-state error, with dynamics given by a simple second-order system response. If we consider a step-input in $v_{p/b}$, corresponding to the person walking at some new self-selected speed, what we will observe is an impulse response of a simple second-order system, which has no steady-state component. Thus, the desired recentering and self-selected-speed components are both achieved.

We can express the characteristic equation of the resulting system in terms of the standard form of a second-order system with damping ratio ζ and natural frequency $\boldsymbol{\omega}_n$ as its parameters: $s^2 + 2\zeta\boldsymbol{\omega}_n s + \boldsymbol{\omega}_n{}^2$.

Such a representation makes the parameter study more intuitive and facilitates a systematic procedure for choosing gains, rather than choosing them on an *ad hoc* basis. Since ζ and $\boldsymbol{\omega}_n$ are positive values, provided our original gains are both positive values, one can easily verify the stability of the system. Both ζ and $\boldsymbol{\omega}_n$ influence the transient response of the controller, which in turn significantly affects the perception of realistic walking on a locomotion interface.

To investigate the effect of parameters on walking with self-selected speed, we simulated the behavior of a user on the belt starting at rest and then changing the self-selected speed to $v_{p/b} = 1$ m/s, and then coming to a stop after a few seconds. In Fig. 3 we see the different behaviors of x_p due to changes in the value of ζ, while $\boldsymbol{\omega}_n$ is held constant. For $\zeta < 1$ (underdamped), the user's position has an oscillatory transient response. Such a behavior is not desirable since it

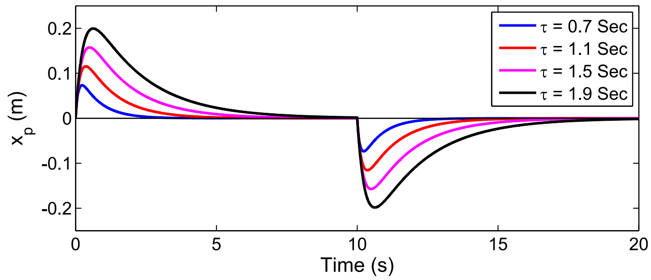


Fig. 4. The effect of different values of τ while $\zeta = 1.5$ on a user's position x_p during walking. Starting from rest, $v_{p/b} = 1$ m/s for the first 10 sec, and then $v_{p/b} = 0$ m/s.

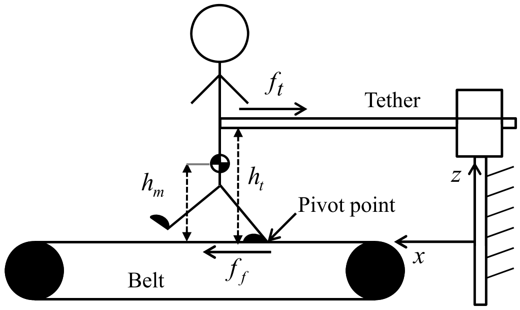


Fig. 5. The tether force is defined as positive when pulling back on the user (in tension). The foot force is defined as positive when the belt is pushing forward on the user's foot (in compression). The tether attachment point is not assumed to be at the user's center of mass in general.

can disturb the user's sense of balance; we find that it is disconcerting to be brought backward on the belt, come to a complete stop relative to the inertial frame, and then be brought forward again. We initially hypothesized that a critically damped system $\zeta = 1$ would be the most desirable, in that it would eliminate any oscillation, but would still result in a fast system. However, we find that a higher damping ratio actually results in a more-desirable response, since the user is not allowed to move forward as much on the belt before being recentered.

An additional benefit of adding more damping to the system is to increase stability robustness. The phase margin of the system is related to the damping ratio [22], with phase margin being quite sensitive to ζ in the approximate range $0 < \zeta \leq 1.2$, and with diminishing returns on phase margin for further increases in ζ. In our pilot tests, we found that increasing $\zeta > 1.5$ did not lead to noticeable differences in the controller; this can also be observed in Fig. 3 by comparing the responses with $\zeta = 1.5$ with $\zeta = 2$, which are very similar. We also note that these system properties were observed regardless of the value of ω_n used (although that would affect the settling time). From the above considerations, we conclude that a damping ratio value of $\zeta = 1.5$ is a desirable value in terms of system response and stability robustness, and we will use this value throughout the remainder of the experiments.

We can now investigate the effect of ω_n on the response of the system. Having selected a constant $\zeta = 1.5$, rather than using ω_n, we can use a more intuitive parameter: an effective time-constant τ. For an overdamped system, the dominant (i.e., slowest) pole is located at

$$r_{\text{dom}} = -\zeta\omega_n + \omega_n\sqrt{\zeta^2 - 1} \tag{5}$$

and the effective time-constant is calculated as

$$\tau = -\frac{1}{r_{\text{dom}}}. \tag{6}$$

Given any desired combination of damping ratio and time-constant, we can compute the required natural frequency:

$$\omega_n = \frac{1}{\tau(\zeta - \sqrt{\zeta^2 - 1})}. \tag{7}$$

Finally, we set our gains as $K_x = \omega_n^2$ and $K_v = 2\zeta\omega_n$.

The choice of τ has a significant impact on the behavior of x_p, as shown in Fig. 4, and on the user's perception of realistic walking. If τ is chosen too small, then the controller returns the user to x_{ref} quickly, and in pilot testing we observed that this can result in the belt feeling too responsive, almost as if it moves before the user was expecting it to move. If τ is selected too large, then a larger deviation in the user's position from x_{ref} is tolerated by the controller, which can result in the user walking nearer to the edge of the belt (and closer to the screens) before being drawn back to the center, which can be disconcerting. The value of τ is limited on the low end by the belt motor's capabilities, and on the high end by the allowable traveling distance from x_{ref} based on the length of the belt and other similar constraints.

2.2 Kinesthetic Force Feedback

When walking overground, a person must put in mechanical work to accelerate their body, equal to the change in kinetic energy of the person's mass, and this work is ultimately done by the person's feet applying forces to the ground. When walking through a virtual world on a treadmill, it is possible to accelerate through the virtual world by simply increasing the belt speed, without the user putting in the same amount of work that would be required overground; this can negatively impact on the user's sense of balance, and can lead to a locomotion interface that feels unstable (similar to walking on ice). The Treadport utilizes a mechanical tether to apply a force f_t to the user's torso, as shown in Fig. 5, and this kinesthetic force feedback can be used to increase the user's sense of balance and stability, and to make the work done by the user to accelerate in the virtual world similar to overground walking.

In prior work with the Treadport [9], the kinesthetic force feedback was set as *inertial force feedback*:

$$f_t = -ma_b, \tag{8}$$

where m is the mass of the user. The rationale behind inertial force feedback is that if a person were accelerating overground with some acceleration a, it would require the person to generate a net forward force equal to ma to cause that acceleration, so the kinesthetic force feedback in the Treadport should demand such a force from the user. This inertial force feedback results in a reasonably good walking experience for the user, but often with a sense that the effective inertia of the user in the virtual environment seems slightly too high, such that starting from rest and coming to rest both seem slightly too difficult. An *ad hoc* tuning parameter to attenuate the force

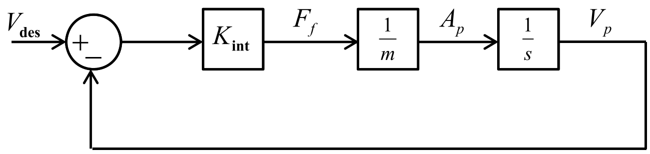


Fig. 6. Simple human walking model.

often results in a more desirable experience for users, and it was hypothesized that this parameter was necessary due to inaccurately modeling the user as a point mass concentrated at the harness contact point rather than considering the whole body [9].

In this section we reconsider the kinesthetic force feedback of the tether, in an attempt to provide a more realistic walking experience. Rather than considering a person walking on a treadmill belt, we instead consider a person standing on a stationary belt, and imagine that the belt is accelerating forward without the user's knowledge. This would feel like the ground was being pulled out from under the user's feet; the inertia of the user's body would not allow the body to accelerate with the feet (we assume no slip between the feet and the belt, due to friction), resulting in an angular rotation of the body in the sagittal plane and a negative impact on the user's balance. However, it is easy to imagine in this scenario that there exists a tether force (pushing forward in this case) that would prevent the user's body rotation and thus prevent the loss of balance (similar to the role played by the handrail on an escalator or moving walkway). Our new *balance-based force feedback* provides such a force.

To calculate the correct tether force f_t for balance-based force feedback, we consider the user on the Treadport as illustrated in Fig. 5. We assume the user to be a rigid body in contact with the ground at some pivot point with a no-slip condition, with a body center of mass at a height h_m, and with a tether applying a force at a height h_t. There is a force f_f that the belt applies to the user's foot (defined positive in the forward direction), but which is unknown to us. Our goal is to set f_t such that the user's center of mass moves forward with the same acceleration as the belt, due to the two applied forces:

$$f_f - f_t = m a_b. \tag{9}$$

We would also like the the resulting moments of the two applied forces to result in no rotation of the user's body about the center of mass in the sagittal plane:

$$f_f h_m + f_t(h_t - h_m) = 0. \tag{10}$$

By combining (9) and (10) to eliminate the unknown force f_f, we solve for the correct tether force:

$$f_t = -\frac{h_m}{h_t} m a_b. \tag{11}$$

The value of h_t can be easily measured for a given user after the user dons the harness. The value of h_m is not trivial to measure, but it can be approximated with good accuracy given only a measurement of a user's height H as $h_m = 0.58H$ [23].

This balance-based force feedback is similar to the previous inertial force feedback in that it is proportional to the user's mass and the belt's acceleration, but typically smaller due to the coefficient h_m/h_t, which is typically less than 1. This result explains the need for the previous *ad hoc* tuning parameter to attenuate the inertial force feedback. Although the new balance-based force feedback was derived using a thought experiment that involved a *standing* user, we will show later in this paper that this new method is preferred by users over the previous inertial force feedback when *walking* on the Treadport under a variety of conditions.

Note that the value of f_t in (11) is the correct value to apply for a given belt acceleration a_b, regardless of how that value was selected or achieved; in this way, the kinesthetic-force-feedback controller is truly independent of the belt controller. However, the stability of the complete system, including the user and the various distributed Treadport controllers, must still be considered.

2.3 Stability Analysis

The most commonly used model for human running is

$$v_p = v_{\text{des}}(1 - e_{\text{int}}^{-t/\tau}), \tag{12}$$

where v_p is a person's speed overground, v_{des} is their desired speed, and τ_{int} is the person's inherent time constant [24]. We make the assumption that this common running model is a reasonable approximation for walking and running on the Treadport. As illustrated in Fig. 6, this observed behavior can be predicted by modeling a person as a mass m that controls their speed using a proportional feedback controller on velocity. The relationship between the person's internal "gain" K_{int} and the resulting time constant τ_{int} is

$$\tau_{\text{int}} = \frac{m}{K_{\text{int}}}. \tag{13}$$

In this simple walking model, the ground applies f_f to the person's foot, which is a reaction force to the propulsive force applied by the person to the ground, to accelerate them with a_p.

Walking on the Treadport is similar to overground walking, with two major differences. First, the net force causing the person's acceleration a_p in the inertial frame is obtained by considering both the force f_f from the belt to the person's foot and the force f_t applied by the tether. Second, the person compares their desired walking speed v_{des} with their speed relative to the belt's speed $v_{p/b}$.

The resulting closed-loop system comprising a user in the Treadport is depicted in Fig. 7; we have included all of the elements that affect the systems dynamics, including the low-level belt-speed controller and the differentiation filter used. To analyze the stability of the system, we convert the equations into the Laplace domain. To be concise, we use $\psi = h_m/h_t$ in the subsequent equations. Note that $\psi = 1$ when the tether applies its force directly at the user's center of mass, and in the Treadport $\psi < 1$ typically.

The belt speed $V_b(s)$ can be expressed as a function of the two exogenous inputs as

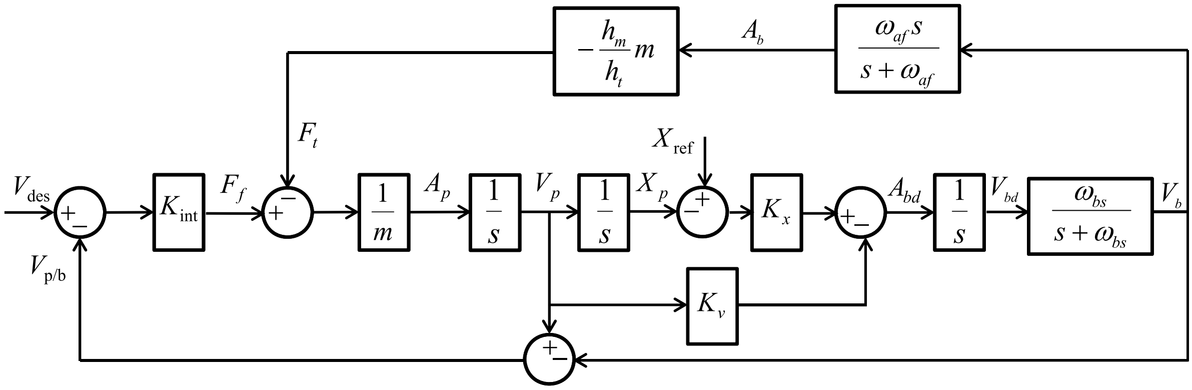


Fig. 7. Block diagram of a user in the Treadport. The user's desired velocity V_{des} is set internally by their own volition. The reference position X_{ref} is set in the control software and is typically constant.

$$V_b(s) = \left(\frac{\omega_{\text{bs}} K_x s(ms + K_{\text{int}})(s + \omega_{\text{af}})}{\Delta(s)}\right) X_{\text{ref}}(s) - \left(\frac{\omega_{\text{bs}} K_{\text{int}}(K_v s + K_x)(s + \omega_{\text{af}})}{\Delta(s)}\right) V_{\text{des}}(s), \tag{14}$$

where $\Delta(s)$ is the system's characteristic equation:

$$\begin{aligned}\Delta(s) = {} & ms^5 + (K_{\text{int}} + \omega_{\text{af}} m + \omega_{\text{bs}} m)s^4 \\ & + (K_{\text{int}}(\omega_{\text{af}} + \omega_{\text{bs}}) + \omega_{\text{bs}}\omega_{\text{af}} m)s^3 \\ & + \omega_{\text{bs}}(K_{\text{int}}(K_v + \omega_{\text{af}}) + \omega_{\text{af}} m\psi)s^2 \\ & + \omega_{\text{bs}}(K_{\text{int}}(K_x + \omega_{\text{af}} K_v) + K_x \omega_{\text{af}} m\psi)s \\ & + \omega_{\text{bs}}\omega_{\text{af}} K_{\text{int}} K_x,\end{aligned} \tag{15}$$

where ω_{af} is the acceleration-differentiator's filter corner frequency, and ω_{bs} is the bandwidth of the low-level belt-speed controller (see supplemental material). Equation (14) indicates that if a user wants to walk with V_{des}, the belt speed asymptotically approaches V_{des}, but in the opposite direction as expected ($V_b \to -V_{\text{des}}$), provided the system is stable.

Something that is not immediately evident from (15) is that the user's mass cancels out of the characteristic equation, once the recentering-controller gains are set as described.

To investigate the stability of the system, we first numerically explored the range of specific values being proposed for the Treadport: $\omega_{\text{af}} = 16$ rad/s, $\omega_{\text{bs}} = 9$ rad/s, $\zeta = 1.5$, $\tau_{\text{int}} \in [0.85\text{-}1.29]$ (based on [24]), $\tau \in [0.7\text{-}1.9]$ (time-constants used in the study), and $\psi \in [0.87\text{-}0.92]$ (the range of values measured with our human subjects in this study); we found that all combinations of values are stable. Next, we conducted a sensitivity analysis by fixing all variables but one at their nominal values (using $\psi = 0.9$, $\tau = 1.26\,s$, and $\tau_{\text{int}} = 1$ as nominal) and then varying the remaining parameter from zero to infinity, using root-locus techniques, and examined the effect on stability. We found the stability ranges of the mentioned parameters to be as follows: $\omega_{\text{af}} \geq 3$ rad/s, $\omega_{\text{bs}} \geq 2$ rad/s, $1 \leq \zeta \leq 2.3$, $\tau_{\text{int}} > 0$, $\tau \geq 0.46$, and $0.1 \leq \psi \leq 3.9$. All of the values used in this study safely fall in the stability ranges of these parameters. It is observed that if no force feedback were to be used (i.e., $\psi = 0$) with all other parameters held constant, the system would become unstable. This observation implies that if no force feedback is used during locomotion on the Treadport, the stability of the system should be ensured by changing other parameters prior to any experiment.

3 Experiment Design and Methods

The purpose of the new controller is to enhance the realism of walking on the Treadport in several ways: the belt's response to a user's motion should be such that it does not harm their perception of realistic walking; a user should be able to start walking and come to a stop without feeling excessive pulling/pushing forces; a user should be able to maintain any reasonable self-selected walking speed similar to overground locomotion; and a user should be able to smoothly transition between forward and backward walking without any modification to the controller.

In order to evaluate our proposed changes to the Treadport controller in light of the desired characteristics described above, we conducted three separate human-subjects experiments. The purpose of the experiments were threefold: First, to find the most preferred value for the recentering-controller time constant τ (which we will denote by τ_p). We conducted an experiment to test the hypothesis that the controller's time constant influences users' perception of realistic walking, and that there exists a most-preferred time constant. Second, to compare the proposed balance-based force feedback with the previous inertial force feedback. We conducted a second experiment to test the hypothesis that the proposed method leads to more realistic walking than the previous method. Third, to quantitatively evaluate the ability of subjects to attain any self-selected walking speed and maintain it. We conducted a final experiment to test the hypothesis that subjects would be able to attain and maintain four distinct self-selected speeds denoted qualitatively as "normal" walking, "fast" walking, "jogging," and "backward" walking.

Our goal is to determine the most preferred recentering-controller time-constant and kinesthetic-force-feedback method independently of one another. However, both controllers must be active for the Treadport to function properly, so it is impossible to completely isolate the effects of the two controllers. Pilot testing provided strong evidence that our new kinesthetic-force-feedback controller was significantly superior to the previous controller, whereas pilot testing for the recentering-controller was not as conclusive.

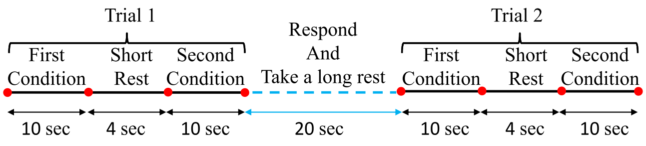


Fig. 8. The timing used within and between trials in the two-alternative-forced-choice experiments.

Therefore, we structured our experiments as follows. First, we conducted our recentering-controller experiment while always using our new kinesthetic-force-feedback controller (which we hypothesize is superior). Then, after finding the preferred recentering-controller, we conducted our kinesthetic-force-feedback experiment to verify that our original hypothesis was correct, and that the new controller was indeed superior to the previous controller.

We chose twenty healthy subjects with a range of height (1.76 ± 0.07 m) and weight (78.87 ± 13.59 kg). Subjects' ages ranged from 19 to 32 years. Subjects were naïve with respect to the experiment. The inclusion criterion was that a participant could fit well in the harness such that mechanical coupling between the participant and the attached tether was maximized. Subjects were provided with written instructions explaining the experiment.

3.1 Recentering-Controller Time-Constant

In pilot testing we determined two extreme values for τ. Enforcing a maximum value of τ prevents a user from getting too close to the Treadport's front edge; this value is set at $\tau = 1.9$ s. Enforcing a minimum value of τ prevents commanding belt motions that are too large to actually be achieved by the Treadport hardware; this value is set at $\tau = 0.7$ s. The full range was then divided into three equal regions by considering four values of τ: $\tau_A = 0.7$ s, $\tau_B = 1.1$ s, $\tau_C = 1.5$ s, and $\tau_D = 1.9$ s.

We designed the experiment based on the two-alternative-forced-choice (2-AFC) method, which is widely used in sensory tests [25]. Within a given trial, subjects were forced to choose between two different conditions, each corresponding to a different τ value unknown to them. Within a given trial, subjects were asked to start walking from rest and then stop walking, and to repeat this process as many times as possible within the time provided. There were no instructions about what their speed should be. They were then asked to simply choose the condition that they preferred. There were no instructions given as to how the subjects should make their determination of preference. We used computerized auditory cues through a speaker to inform subjects about: the trial's number; the beginning and the end of each trial; and the condition's number (i.e., "first" or "second"). After completion of each trial, subjects were prompted to select their preferred condition by saying either "first" or "second."

For the first part of the experiment, there were six possible combinations for all τ pairs to present to the subjects using the 2-AFC method: {(τ_A,τ_B), (τ_A,τ_C), (τ_A,τ_D), (τ_B,τ_C), (τ_B,τ_D), (τ_C,τ_D)}. We presented each pair twice to improve the power of the experiment, resulting in 12 total trials per subject. The order of these twelve trials was fully randomized, as was the ordering of the two τ values presented within a given trial. Fig. 8 shows the timing of the experiment within and between trials. Before starting the experiment, subjects had a one-minute period for familiarization with the Treadport.

From the results of the first part of the experiment, we chose the two most preferred τ values (i.e., the two values selected most often). In the event that three τ values were selected equally, we planned to perform an additional six trials (three combinations with two repetitions) to narrow the selection down to the two most preferred values, but this eventuality never occurred in our experiment.

In a second part of the experiment, immediately following the first part, we presented the subject with their two most preferred τ values in six repeated trials, with the order of the conditions within each trial fully randomized. We again used the timing shown in Fig. 8. Again, the subjects were asked to state their preference. A value of six trials was chosen because, when using the 2-AFC method, six is the minimum number required such that if the subject chooses the same condition for all trials we can say with 95 percent certainty that they prefer that condition. At the end of the experiment, we asked the subjects to fill out a questionnaire comprising a single question: "When selecting the condition that you preferred in a trial, what was your preference based on?"

We utilized the convex combination of the results to estimate the preferred time-constant τ_p for each subject, where the weighting coefficients were the fraction of the times that each of the two τ values were selected in the second part of the experiment. For example, if τ_B was chosen four out of six trials, and τ_C was chosen two out of six trials, then τ_p would be calculated as $\tau_p = (4/6)\tau_B + (2/6)\tau_C$. This method essentially performs an interpolation between tested values, with the assumption that there exists an underlying continuous preference function with a local maximum value.

3.2 Kinesthetic-Force-Feedback Method

This experiment was carried out after completion of the experiment of Section 3.1, on a different day. Participants were presented with the two different force-feedback methods through 12 trials, again using the 2-AFC method, with the order of the conditions randomized within trials. The instructions provided to the subjects were identical to those of the experiment of Section 3.1, and the timing within and between trials is again depicted in Fig. 8. With 12 trials, a given subject must choose a condition at least 10 times out of 12 in order to say that the subject prefers the condition with 95 percent confidence. In this entire experiment, the time-constant of the recentering controller was set at the mean value of τ_p across all twenty subjects, obtained as described in Sections 3.1.

3.3 Ability to Walk at Self-Selected Speeds

The final experiment was conducted immediately following the experiment of Section 3.2. We asked the subjects to walk with various speeds and to maintain their speed for a given period of time. Four qualitative walking speeds were used: "normal" walking, "fast" walking, "jogging," and "backward" walking. The subjects were not provided with any quantitative definition of these terms, and were asked to self-select the speed that best represented the qualitative

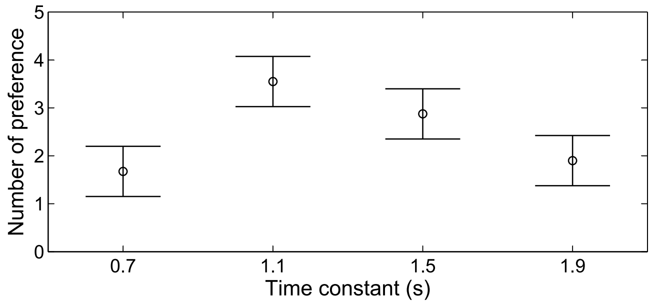


Fig. 9. Preference mean with 95 percent confidence interval for each belt-controller time-constant across 20 subjects.

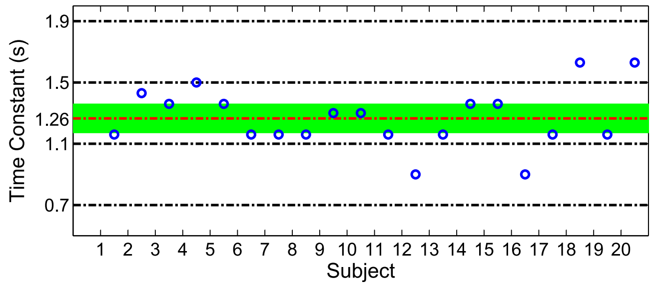


Fig. 10. Subjects' individual preferred belt-controller time-constants τ_p are given with circles. The mean preferred time-constant across subjects is at $\tau_p = 1.26$ s, and the 95 percent confidence interval is shown as a bar.

terms. For normal walking, we instructed the subjects to walk at their preferred speed as if they were walking down a hallway. For fast walking, we instructed the subjects to imagine that they were walking down a hallway in a hurry, but to not run. Subjects did not receive any explicit guidance for jogging or backward walking.

For a given subject, the order of the four walking speeds was randomized. The subject was given a verbal instruction of which walking speed they would be attempting, with the instruction to walk at that speed, starting from rest, until they were informed of the end of the trial, at which point they should come to a stop. Before each trial, the subject spent 20 s to practice their assigned walking speed. We then used computerized auditory cues through a speaker to inform subjects about the beginning and the end of each trial. Each trial lasted for 20 s. We continuously recorded the belt's speed during the trials. The wait time between trials was approximately 20 s, but varied from trial to trial and between subjects.

To quantify the ability to maintain a given speed, we consider the standard deviation σ and mean μ of the belt's speed for the final 15 s of data (thus removing the transient effects observed in the first 5 s of data). We use the coefficient of variance in speed when attempting to walk at a constant speed as the quantitative measure for evaluating the Treadport's performance:

$$C_v = \frac{\sigma}{\mu}. \tag{16}$$

The importance of minimizing variance in self-selected speed has been considered previously [13], [26]. The lower the C_v, the easier it is to maintain a constant speed.

4 Experiment Results

4.1 Recentering-Controller Time-Constant

Throughout the twelve trials in the first part of the experiment, the number of times that a given τ could be preferred could vary from zero (i.e., never preferred) to six (i.e., always preferred). Fig. 9 depicts that in the first part of the experiment, subjects chose the conditions corresponding to $\tau = 1.1$ s and $\tau = 1.5$ s most often as their preference. Since our results were non-parametric, we used Friedman's test and Dunn-Sidak *post hoc* analysis for multiple comparison [27]. We find a statistically significant difference between either of the two most-preferred time constants and either of the two least-preferred time-constants, but we do not find a statistically significant difference between the two most-preferred time-constants.

The second part of the experiment used the two most-preferred time-constants for each individual subject to determine a more accurate value of the preferred τ_p for that subject. The results of the second part of the experiment are presented in Fig. 10. The mean preferred time-constant across all subjects, with a 95 percent confidence interval, is $\tau_p = 1.26 \pm 0.09$ s. 16 out of 20 subjects had an individual τ_p in the range 1.1-1.5 s, which is sufficient to say with 95 percent confidence that the entire population as a whole will prefer a τ in this range. In looking at Fig. 4, we believe this is a nonobvious result.

We also investigated the effect of subjects' height, weight, self-selected speed, and the tether attachment point encoded by ψ (which is correlated with height) on their preferred time-constant. Statistical analysis using linear regression revealed no statistically significant effect of any of these four parameters on the preferred time-constant. It can be inferred that despite their different heights and weights, the subjects had a similar walking preference on the Treadport, and variance between subjects is likely due to personal preference as opposed to a quantifiable anatomical characteristic.

Based on the results of this experiment, we conclude that it is reasonable to use a time-constant of $\tau = 1.26$ s for all users in the future, and no additional user-specific measurements (i.e., height or weight) can be used to improve the value of τ. Any additional user-specific improvements to τ would essentially require this experiment to be recreated on a user-by-user basis. Fortunately, we see in Fig. 10 that the variance between users is relatively small, and most users will be satisfied with $\tau = 1.26$ s.

To see how the subjects perceived the effect of different τ's, we used a questionnaire at the end of the experiment. Their responses to our question about what their preference was based on revealed that subjects typically determined one condition to be inferior to the other condition, and then voted *against* that condition (as opposed to voting *for* the other). Comments typically took one of three forms: (1) Sometimes the belt was too responsive and started moving sooner or faster than I expected. (2) Sometimes the belt was too sluggish and it seemed to take too much effort to accelerate or decelerate. (3) Sometimes the belt seemed to keep moving for too long after I tried to stop walking. From our own experience in pilot testing, we know that the first response is due to τ being too small, and the second and third responses are due to τ being too large.

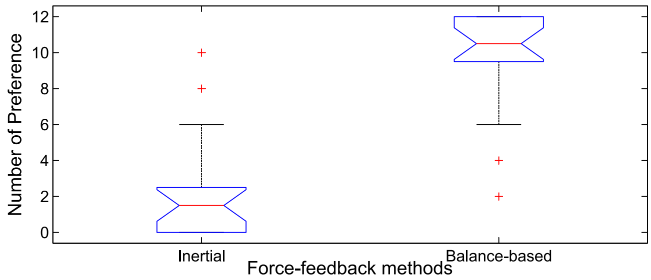


Fig. 11. Preference of force-feedback methods, shown as a notched-box-whisker plot.

4.2 Kinesthetic-Force-Feedback Method

We compared the previous inertial-force-feedback method to the new balance-based-force-feedback method using the 2-AFC method. Fig. 11 indicates that the new method is significantly preferred across the 20 subjects (and thus the population). A binomial distribution is the appropriate way to analyze statistical significance of 2-AFC tests [28], and it enables us to analyze each subject's preference individually.

To state with 95 percent certainty that the new method is preferred over the previous method, we need to show that there is less than a 5 percent chance that random guessing could have led to the number of preferences of the new method. The probability of choosing the new method in a trial merely by guessing is $p = 0.5$ (there are only two choices). For the n trials of any given subject, the null hypothesis is that the two methods are identical, which means the chance of choosing the new method out of n trials is equal to the chance of choosing the previous method. The alternative hypothesis is that the new method performs differently than the previous method, so that the difference observed in the preferences was not obtained by guessing. We need to choose the new method in at least T trials out of n trials in order to conclude that the new method is selected significantly more often than it would be by chance. The probability of choosing the new method at least T times out of n just by chance (denoted by $P(X \geq T)$) should be less than α ($\alpha = 0.05$ for 95 percent confidence). Thus:

$$P(X \geq T) = (1 - P(X \leq T - 1)) \leq \alpha, \tag{17}$$

where the cumulative distribution function of a binomial distribution can be expressed as:

$$P(X \leq T - 1) = \sum_{i=0}^{T-1} \binom{n}{i} p^i (1-p)^{n-i}. \tag{18}$$

With $p = 0.5$, $\alpha = 0.05$, and $n = 12$ (i.e., 12 trials for a given subject), we calculate $T = 10$, meaning that the new method must be preferred in at least 10 out of 12 trials to be 95 percent confident in the preference. Fifteen subjects chose the new method in at least 10 out of 12 trials. One subject chose the new method with 90 percent confidence, and three subjects did not have any significant preference. Only one subject preferred the previous method with 95 percent confidence. The fact that 15 out of 20 subjects preferred the new method with 95 percent confidence is sufficient for us to also conclude that the *population* will prefer the new method with 95 percent confidence. That is, if we consider $p = 0.5$, $\alpha = 0.05$, and $n = 20$ (i.e., the total number of subjects), we calculate that $T = 15$. This confirms the results shown in Fig. 11.

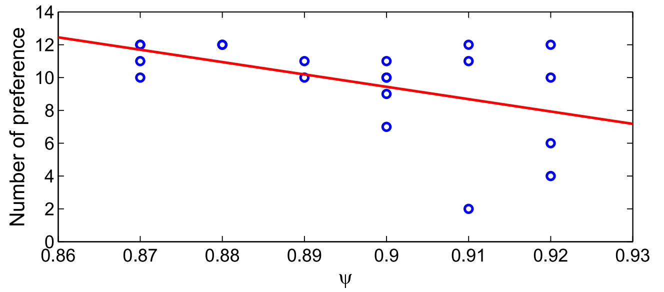


Fig. 12. Preference of balanced-based force feedback (out of a possible 12) as a function of ψ. Note that there are two data points at each of (0.87,12), (0.88,12), and (0.9,10). The effect of ψ is statistically significant.

We considered the effect of the tether attachment point relative to the center of mass of the user, through the variable ψ, on the preference of the balance-based force feedback over inertial force feedback. The results are shown in Fig. 12. We find that there is a statistically significant effect of ψ on preference, with lower values of ψ (i.e., the tether being attached farther above the user's center of mass) resulting in more likely preference for the balance-based method. We will return to this result in Section 5.

In the post-experiment questionnaire, the subjects mentioned several reasons for their preference. Again, they tended to vote *against* one method, rather than voting *for* the other. Comments typically took one of three forms: (1) Sometimes I was too aware of the tether pushing/pulling on my back. (2) Sometimes it required too much effort to get the belt to move. (3) Sometimes my balance was disturbed when I tried to come to a stop. These comments echo our own observations that led us to reconsider the kinethetic-force-feedback method in the first place.

4.3 Ability to Walk at Self-Selected Speeds

In the final experiment, we used each subject's preferred kinesthetic-force-feedback method (whether or not it was significantly different from the other method), and used $\tau = 1.26$ s for all subjects. Fig. 13 depicts the self-selected speeds of each subject for each of the four qualitative speed types. Each of the means and standard deviations shown are the result of experimental data of the type shown in Fig. 14, after removing the transient effects of the first 5 s of data. We can readily see that subjects are able to achieve self-selected speeds and maintain those speeds. We see that the user's perceived walking speed $v_{p/b}$ has more variance than the speed of the belt itself, indicating the periodic (rather than constant) walking speed reminiscent of natural walking. We also observe that the speed profiles are in agreement with the standard first-order model [24].

The mean (with 95 percent confidence interval) of C_v values from (16) for the four different qualitative speeds are presented in Fig. 15. It can be observed that C_v is less than 5 percent for normal walking, fast walking, and jogging, and there is no significant difference between them. Although C_v of backward walking is significantly larger than the other three cases, it is still relatively small. Backward walking seems to be not as intuitive as forward

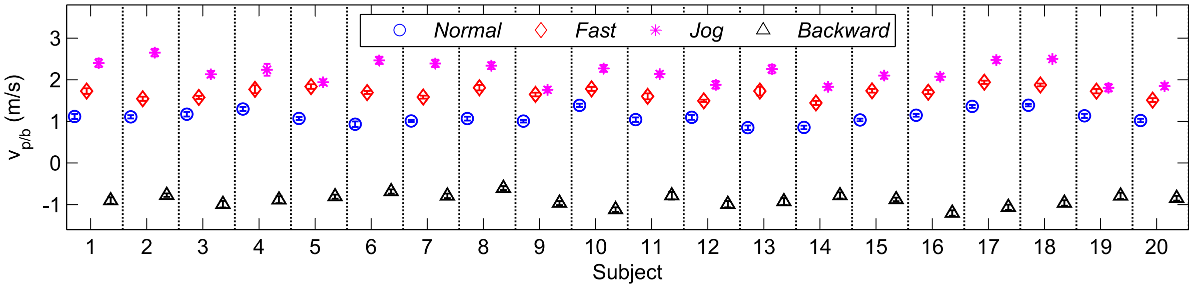


Fig. 13. Subjects' mean walking speed with standard deviation for the four qualitative speeds.

walking, so this result is not surprising; we observe that users usually have more difficulty in keeping a constant pace. Thus, it is critical that a locomotion interface provides a stable and safe condition for a user to experience backward walking.

Walking on the Treadport is similar to overground walking. All subjects could perform all of the four qualitative walking types, as well as starting from rest, coming to stop, and standing still, with smooth transitions between these modes. Fig. 16 demonstrates all of the above mentioned tasks carried out on the Treadport for a typical user. The ability to come to a stop and stand still can be observed around 60 s. A quick transition from backward walking to forward walking can be observed around 90 s. We plot $-v_b$ rather than $v_{p/b}$ so that the perfectly motionless condition can be clearly seen during the standing-still portions ($v_{p/b}$ exhibits the user's natural sway).

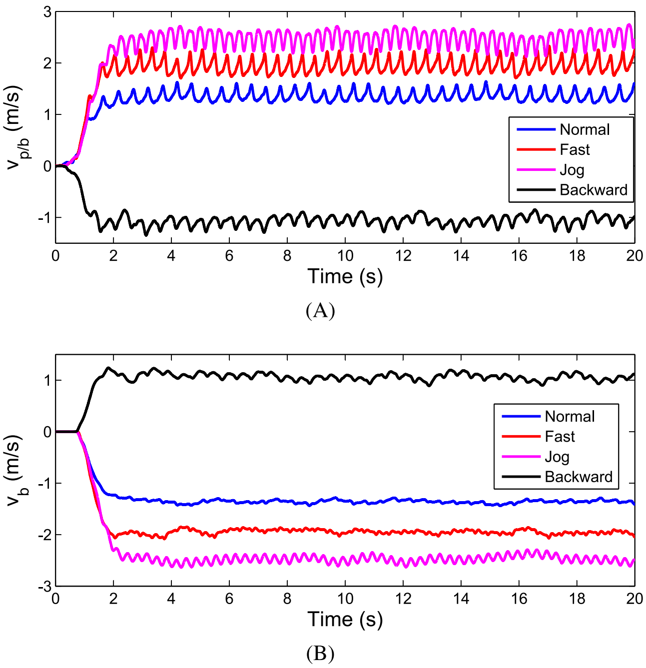


Fig. 14. Four different self-selected qualitative speeds for a typical user. (A) Walking speed of the user relative to the belt $v_{p/b}$. (B) The belt's speed v_b.

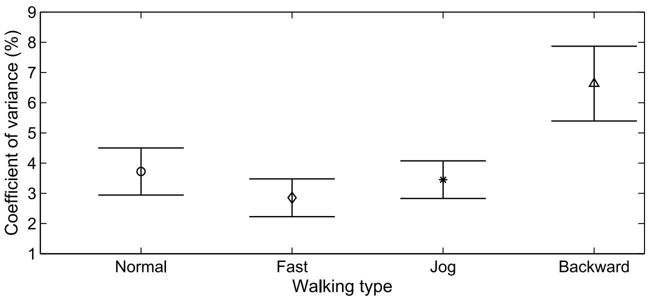


Fig. 15. Mean value of C_v with 95 percent confidence interval for different walking types.

5 Discussion

For further analysis of the new controller, we consider the energy expended by a user to change their walking speed. This expended energy can be thought of as the work done by the user's feet when pushing the ground away, or alternatively, as the work performed on the user's center of mass (COM) by the ground. The relationship used for COM work is given by [29]:

$$w = \int f_f \cdot v_{COM} dt, \tag{19}$$

where $w(t)$ is equal to the user's expended energy as a function of time, f_f is the force applied between the ground or belt and the user's feet (positive in compression), and v_{COM} is the speed of the user's COM. In overground walking, $v_{COM} = v_p$. In the Treadport, $v_{COM} = v_{p/b}$ [21]. Using numerical simulations, the value of $w(t)$ in Treadport walking is evaluated by using the block diagram in Fig. 7 for different values of $\boldsymbol{\psi}$, and $w(t)$ in overground walking is evaluated from the block diagram of Fig. 6. Fig. 17 depicts the energy expenditure $w(t)$ for a typical human with the mass of 80 kg, starting from rest and reaching a steady-state walking speed of 2 m/s.

The results indicate that using both inertial and balance-based force feedback results in energy expenditure on the Treadport that is similar to overground. However, inertial force feedback results in more energy expenditure than overground, whereas balance-based force feedback for all the values of $\boldsymbol{\psi}$ used in this study results in less energy

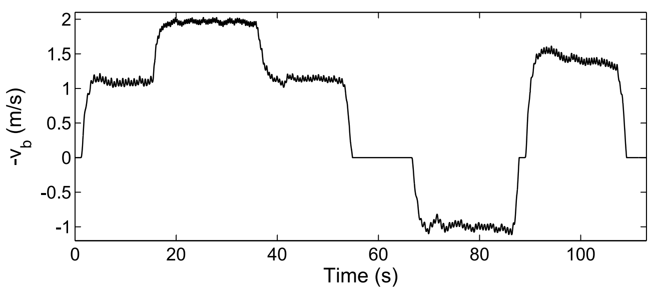


Fig. 16. Walking speed while transitioning between different walking types for a typical user.

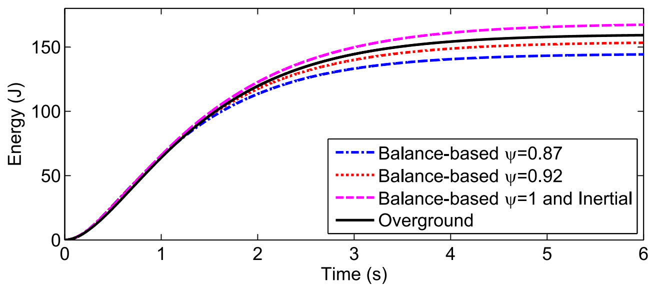


Fig. 17. Energy expenditure in overground walking and the Treadport walking of a typical user during different force-feedback methods.

expenditure than overground. This trend is independent of the user's mass and walking speed. For the preferred time-constant and the full range of ψ from our studies, users would expend 5.3 percent more energy than overground with inertial force feedback, whereas they would expend 3.7-9.4 percent less energy than overground with the balance-based force feedback, depending on ψ. It is possible to modify the tether attachment point to result in a perfect energy expenditure when using balance-based force feedback ($\psi = 0.95$, for the recentering-controller parameters used here, independent of user mass and steady-state speed). However, with inertial force feedback, even if we consider modifying the recentering-controller time-constant over the range of reasonable values found in our study, the inertial force feedback would always result in 5.0-5.5 percent more energy expenditure than overground.

Regardless of the tether attachment point, once ψ is set, the balance-based force controller is the correct force in terms of the user's sense of balance, and we believe that this is the primary reason for subjects' preference of the method. The improvement in balance and feeling of safety is particularly noticeable when coming to a stop, during the brief period after the feet are firmly planted on the belt but before the belt has come to a complete stop. The improvement is also noticeable when starting from rest, removing the need of the user to lean forward slightly during the acceleration phase. It seems likely that energy expenditure is also contributing to the subjects' preference, with users preferring to expend less energy. The dependence of user preference on ψ, seen in Fig. 12, provides evidence of this dependence, in light of the results shown in Fig. 17. Even if energy expenditure were the only factor determining user preference, the balance-based method would still be the preferred method with $\psi = 0.95$, since the inertial method would require 5.3 percent more energy.

6 Conclusion

In this paper we described an improved control system for the Treadport immersive locomotion interface, with results that should generalize to any treadmill that utilizes an actuated tether to enable self-selected walking speed. A new controller for the belt was implemented to regulate the user's position to some reference position (typically near the center); when combined with the user's volition, this same controller also enables the user to naturally self-select their walking speed as they would when walking over ground. We found that a simple proportional-derivative controller is effective at regulating the user's position, and it is most natural feeling when it has a damping ratio of approximately 1.5 and a time constant of approximately 1.26 s. A new kinesthetic-force-feedback controller was designed for the tether that applies forces to the user's torso. This new controller was derived based on maintaining the user's sense of balance during belt acceleration, rather than by rendering an inertial force as was done in prior work. Based on the results of our human-subjects study, both the belt controller and the kinesthetic-force-feedback controller significantly contribute to an improved perception of realistic walking on the Treadport. Our improved controllers use intuitive dynamic-system and anatomical parameters, rather than gains that require *ad hoc* tuning. Our controller simply requires three measurements to be made for a given user—the user's mass, the user's height, and the height of the tether attachment point on the user's torso—and all other controller parameters can be set constant for the entire population.

Acknowledgments

The authors would like to thank Ed Colgate, Robert Riener, and Mark Minor for their constructive comments that led to this redesign of the Treadport control system, as well as the anonymous reviewers for their helpful suggestions. This work was supported by the National Science Foundation through grant 1208637. Babak Hejrati is the corresponding author of the article.

References

[1] V. Dietz and S. Harkema, "Locomotor activity in spinal cord-injury persons," *J. Appl. Physiology*, vol. 96, pp. 1954–1960, 2004.
[2] A. L. Hicks, "Treadmill training after spinal cord injury: It is not just about the walking," *J. Rehabil. Res. Develop.*, vol. 45, no. 2, pp. 241–248, 2008.
[3] R. Boian, G. Burdea, and J. Deutsch, "Robotics and virtual reality applications in mobility rehabilitation," in *Rehabilitation Robotics*, S. S. Kommu, Ed. Vienna, Austria: I-Tech Educ. and Publishing, 2007, pp. 27–42.
[4] J. M. Hollerbach, Y. Xu, R. Christensen, and S. C. Jacobsen, "Design specifications for the second generation Sarcos Treadport locomotion interface," in *Haptic Symposium, Proc. ASME Dynam. Syst. Control Division*, 2000, vol. 69, pp. 1293–1298.
[5] D. Grow and J. M. Hollerbach, "Harness design and coupling stiffness for two-axis torso haptics," in *Proc. Symp. Haptic Interfaces Virtual Environments Teleoperator Syst.*, 2006, pp. 83–88.
[6] C. Cruz-Neira, D. Sandin, T. A. DeFanti, R. V. Kenyon, and J. C. Hart, "The cave audio visual experience automatic virtual environment," *Commun. ACM*, vol. 35, pp. 67–72, 1992.
[7] A. Vijayakar and J. M. Hollerbach, "Effect of turning strategy on maneuvering ability using the treadport locomotion interface," *Presence: Teleoperators Virtual Environments*, vol. 11, no. 3, pp. 247–258, 2002.
[8] S. Kulkarni, C. Fisher, E. R. Pardyjak, M. A. Minor, and J. M. Hollerbach, "Wind display device for locomotion interface in a virtual environment," in *Proc. World Haptics Conf.*, 2009, pp. 184–189.
[9] R. Christensen, J. Hollerbach, Y. Xu, and S. Meek, "Inertial force feedback for the treadport locomotion interface," *Presence: Teleoperators Virtual Environments*, vol. 9, no. 3, pp. 1–14, 2000.
[10] B. Hejrati, D. Hull, J. Black, J. J. Abbott, and J. M. Hollerbach, "Investigation of the treadport for gait rehabilitation of spinal cord injury," in *Proc. Int. Conf. IEEE Eng. Med. Biol. Soc.*, 2012, pp. 4553–4558.
[11] L. Lichtenstein, J. Barabas, R. L. Woods, and E. Peli, "A feedback-controlled interface for treadmill locomotion in virtual environments," *ACM Trans. Appl. Percep.*, vol. 4, no. 1, pp. 1–7, 2007.
[12] J. L. Souman, P. R. Giordona, I. Frissen, A. D. Luca, and M. O. Ernst, "Making virtual walking real: Perceptual evaluation of a new treadmill control algorithm," *ACM Trans. Appl. Percep.*, vol. 7, no. 11, pp. 1–14, 2010.

[13] J. von Zitzewitz, M. Bernhardt, and R. Riener, “A novel method for automatic treadmill speed adaptation,” *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 15, no. 3, pp. 401–409, Sep. 2007.

[14] H. Noma and T. Miyasato, “Design for locomotion interface in a large scale virtual environment. ATLAS: ATR locomotion interface for active self motion,” in *Proc. ASME Dynam. Syst. Control Division*, vol. 64, pp. 111–118, 1988.

[15] J. Feasel, M. C. Whitton, S. Member, L. Kassler, S. Member, F. P. Brooks, L. Fellow, and M. D. Lewek, “The integrated virtual environment rehabilitation treadmill system,” *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 19, no. 3, pp. 290–297, Jun. 2011.

[16] A. Koenig, S. Member, C. Binder, J. Zitzewitz, X. Omlin, M. Bolliger, and R. Riener, “Voluntary gait speed adaptation for robot-assisted treadmill training,” in *Proc. Int. Conf. Rehabil. Robotics*, 2009, pp. 419–424.

[17] J. Yoon, B. Novandy, C. H. Yoon, and K. J. Park, “A 6-dof gait rehabilitation robot with upper and lower limb connections that allows walking velocity updates on various terrains,” *IEEE Trans. Mechatronics*, vol. 15, no. 2, pp. 201–215, Apr. 2010.

[18] J. Kim, H. Park, and D. L. Damiano, “An interactive treadmill under a novel control scheme for simulating overground walking by reducing anomalous force,” *IEEE/ASME Trans. Mechatronics*, doi: 10.1109/TMECH.2014.2341039, pp. 1–6, 2014.

[19] B. A. Frishberg, “An analysis of overground and treadmill sprinting,” *Med. Sci. Sports Exercise*, vol. 15, no. 6, pp. 478–485, 1983.

[20] S. J. Lee and J. Hidler, “Biomechanics of overground vs. treadmill walking in healthy individuals,” *J. Appl. Physiology*, vol. 104, no. 3, pp. 747–755, 2008.

[21] A. Crétual and N. Fusco, “Additional energetic cost due to belt speed variations when walking on a treadmill,” *J. Electromyography Kinesiology*, vol. 21, pp. 551–556, 2011.

[22] K. Ogata, *Modern Control Engineering*. Englewood Cliffs, NJ, USA: Prentice Hall, 2009.

[23] J. H. G. K. S. N. W. D. G. E. Robertson, G. E. Caldwell, *Research Methods in Biomechanics*. Champaign, IL, USA: Human Kinetics, 2004.

[24] A. V. Hill, *Muscular Movement in Man*. New York, NY, USA: McGraw-Hill, 1927.

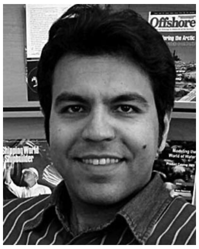
[25] M. Meilgaard, G. V. Civille, and B. T. Carr, *Sensory Evaluation Techniques*, L. Franco, Ed. Boca Raton, FL, USA: CRC Press, 1999.

[26] J. Kim, C. J. Stanley, L. A. Curatalo, and H. S. Park, “A user-driven treadmill control scheme for simulating overground locomotion,” in *Proc. Int. Conf. IEEE Eng. Med. Biol. Soc*,, 2012, pp. 3061–3064.

[27] D. M. F. H. J. Derrac, S. Garcia, “A practical tutorial on the use of nonparametric statistical tests as a methodology for comparing evolutionary and swarm intelligence algorithms,” *Swarm Evol. Comput.*, vol. 1, pp. 3–18, 2011.

[28] J. M. Ennis and V. Jesionka, “The power of sensory discrimination methods revisited,” *J. Sensory Stud.*, vol. 26, pp. 371–382, 2011.

[29] L. H. Sloot, M. M. van der Krogt, and J. Harlaar, “Energy exchange between subject and belt during treadmill walking,” *J. Biomechanics*, vol. 47, pp. 1510–1513, 2014.



Babak Hejrati received the BS degree from Iran University of Science and Technology, Tehran, Iran, in 2006, and the MS degree from K. N. Toosi University of Technology, Tehran, in 2009, both in mechanical engineering. He is currently working toward the PhD degree in mechanical engineering at the University of Utah, Salt Lake City, UT. He is particularly interested in applications of robotic devices for rehabilitation. He is a student member of the IEEE.

Kyle L. Crandall received the BS and MS degrees in mechanical engineering from the University of Utah, Salt Lake City, UT, in 2014. He is currently working toward the PhD degree at the Department of Mechanical and Aerospace engineering in George Washington University, Washington, DC.

John M. Hollerbach received the BS degree in chemistry and the MS degree in mathematics from the University of Michigan, Ann Arbor, MI, in 1968 and 1969, respectively, and the SM and PhD degrees in computer science from the Massachusetts Institute of Technology (MIT), Cambridge, MA, in 1975 and 1978, respectively. He is currently a professor with the School of Computing, University of Utah, Salt Lake City, UT, and an editor of the *International Journal of Robotics Research*. His current research interests include robotics, virtual reality, and human motor control. He is a fellow of the IEEE.

Jake J. Abbott received the BS degree from Utah State University, Logan, UT, in 1999, the MS degree from the University of Utah, Salt Lake City, UT, in 2001, and the PhD degree from Johns Hopkins University, Baltimore, MD, in 2005, all in mechanical engineering. He was then a postdoctoral researcher at ETH Zurich, Switzerland. In 2008, he joined the Department of Mechanical Engineering, University of Utah, Salt Lake City, where he is currently an associate professor. He received the NSF Faculty Early Career Development Award, as well as a number of Best Paper/Poster Awards. He is a member of the IEEE.