Torso Force Feedback Realistically Simulates Slope on Treadmill-Style Locomotion Interfaces

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Abstract

This paper investigates whether torso force feedback on treadmill-style locomotion interfaces can substitute for treadmill tilt to simulate walking on smooth inclines. The experimental platform is the Sarcos Treadport, whose active mechanical tether can apply horizontal forces to the user to simulate the gravity forces experienced in slope walking. The authors show that users are extremely sensitive to slope while walking, being able to discriminate a 0.5 degree slope change. Comparisons are then made between walking on a tilted treadmill platform versus walking on a level platform but with tether force application. Psychophysical experiments show that users select tether forces that are predicted by the gravity forces, although at a 65% fractional force level. These results demonstrate definitively that torso force feedback can realistically simulate gravity forces during smooth slope walking.

KEY WORDS—haptic interfaces, virtual reality, locomotion interfaces, gait, slope walking

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1. Introduction

An ideal virtual reality system would permit completely natural human motions and interactions, such as walking around and manipulating objects. Virtual manipulation is performed with haptic interfaces, a by now well-established field of robotics whose origins are traced to the master hand controllers of teleoperator systems (Sheridan 1992). Virtual walking is afforded by locomotion interfaces, which are often based on exercise machines such as specialized treadmills, 3-D stair steppers, and powered bicycles (Hollerbach 2002). Compared to haptic interfaces, the field of locomotion interfaces is at an early stage that can be characterized as an exploration of the design space of possible devices. They can also be considered as robots because of programmable motion and force application to the feet, especially for the 3-D stair steppers, where the term *foot haptics* has been applied (Iwata 2000b). These 3-D stair steppers are pairs of three-degree-of-freedom (DOF) robots attached to the feet, either in a serial link arrangement such as the Sarcos Biport or a parallel link arrangement such as the Gaitmaster (Iwata 2000a).

The fields of locomotion and haptic interfaces are proceeding more or less independently, so that we cannot yet manipulate while walking. The problem is the small workspace of haptic interfaces versus the large workspace of locomotion interfaces. One solution is some form of portable haptic interface. An example of a portable point force source is the cable-driven HapticGEAR (Hirose et al. 1999), while an example of a portable grasp force source is Virtual Technologies' GraspPack with the CyberGrasp.

As part of certain locomotion interfaces such as the Sarcos Treadport, forces are applied to the trunk of the body to simulate such effects as hitting walls, inertial forces, and slope walking. These forces are applied by active mechanical linkages (i.e., robots) attached to the treadmill frames. We adopt the term *torso force feedback* or *torso haptics* to describe this method of force application. Although the Greek-originated term *haptics* pertains to the hand, its current usage is now so generalized as to pertain to force application to any part of the body (Durlach and Mavor 1994).

Haptic rendering is the method of presenting the interaction forces of contact such as surface hardness, texture, and friction. Haptic rendering takes into account control stability, computational burden, and physiological response properties to fashion engineeringly feasible illusions of manual contact and manipulation. The effects that locomotion rendering seeks to implement include realistic forward motion, turning, slopes and uneven terrain, and walking into obstacles.

Among the proposed designs, those based on treadmills seem currently to offer the best mechanical display for locomotion rendering. Linear treadmills are ubiquitous devices employed commonly for exercise and therefore are a good basis for the proliferation of locomotion interfaces. One can move naturally over the treadmill belt surface, and advanced treadmill designs address reasonably well the variety of desired locomotion rendering effects. The main alternative, the 3-D stair steppers, does not yet allow the freedom of motion possible on treadmills. The design of treadmills for locomotion interfaces has proceeded by adding mechanical complexity either to the treadmill platform or to torso force feedback.

1.1. Mechanical Enhancements of the Treadmill Platform

Commercial linear treadmills are typically equipped with a tilting function that accurately simulates walking up or down a smooth incline. The tilting mechanism is usually slow, though, and does not permit rapid changes in slope to be presented. For example, the original Sarcos Treadport employed a commercial 4-by-8 foot treadmill whose tilting speed was only 1 degree per second (see Fig. 1). The second-generation Treadport involved a special-purpose tilt mechanism designed for a tilting speed of 20 degrees in 1 second for a 6-by-10 foot belt surface (Hollerbach et al. 2000). Figure 2 shows the tilt capability of the new Treadport.

ATR's ATLAS system generalizes the tilting function by placing the treadmill platform on an active spherical joint (Noma and Miyasato 1998). Besides tilting for up and down

Fig. 1. The original Sarcos Treadport.

Fig. 2. The maximum tilt angle of the new Treadport.

slopes, the platform can be tilted sideways to simulate side-slope walking (i.e., walking perpendicular to the fall line). The spherical joint also acts as a turntable to display turning, by swiveling the platform left or right according to a user's intended walking direction. The walking direction is deduced from the amount of lateral motion in a user's step, measured optically by markers on the feet. The belt area is 1.45 m long by 0.55 m wide, the maximum belt speed is 4 m/s, and the maximum turntable rate is 1 rad/s.

Another innovation at ATR is to simulate uneven terrain by deforming a belt with six vertical stages. This Ground Surface Simulator (GSS) can present a slope of up to 5 degrees (Noma, Sugihara, and Miyasato 2000). Because the geometry of the belt changes when deformed by the stages, an active belt tensioning system is employed. The belt dimensions are about the same as for the ATLAS system. Future plans call for mounting the GSS on a motion stage as for the ATLAS system.

Two-dimensional treadmill belts have been devised whose main feature is to allow walking in any direction on the belt surface. The Omni-Directional Treadmill (Darken and Cockayne 1997) consists of a main belt made of rollers, which are made to spin in an orthogonal direction to that of the main belt by another belt underneath. The active surface is 1.3 m by 1.3 m, and the maximum velocity is 3 m/sec.

The Torus Treadmill (Iwata and Yoshida 1999) consists of a main belt fashioned from 12 small treadmills connected side by side. As the main belt moves, the small treadmill belts move orthogonally to create a two-dimensional motion. The Torus Treadmill has a walking area of 2 m by 1.8 m and permits walking speeds of 1.2 m/s.

1.2. Torso Force Feedback

A second line of research adds torso force feedback to simulate locomotion effects. The Sarcos Treadport uses an active tether mechanism that both senses user position and applies forces to the user's back via a torso harness (see Fig. 3). The tether's kinematic structure is a universal joint at the base, a linear joint from the base to the user, and a spherical joint at the user's back. The five rotary joints are passive, but the linear joint is actuated by a geared electric drive acting on a timing belt (Hollerbach et al. 2000). Consequently, the boom can push and pull on the user along a single linear axis, with a force up to 315 N. The resulting torso force feedback has been used for a variety of purposes.

One purpose is to supply an artificial inertial force. In the biomechanics literature, the equivalence of treadmill locomotion to ground locomotion has been debated. Some studies involved constant velocities with no slope (Dal Monte, Fucci, and Manoni 1973; Elliot and Blanskby 1976; Ingen Schenau 1980; Nelson et al. 1972; Nigg, Deboer, and Fisher 1995). There is no consensus on the differences: some researchers found trends among their subjects, while others found significant individual differences but no consistent trends. Ingen Schenau (1980) mathematically showed that at constant velocities, an ideal treadmill and overground running are mechanically the same. He theorized that the observed biomechanical differences might be caused by different surface types, underpowered treadmills that cannot maintain a constant velocity, lack of visual information on the treadmill, or fear of falling off the treadmill.

Fig. 3. The Sarcos Treadport with tether attachment to a user.

With regard to surface type, there is some compliance of the treadmill surface due to the wood backing and its support. McMahon (1984) showed that the stance time during running is affected by surface compliance, and this observation led to the first design of a tuned track with a compliance that minimizes the stance time. The effect for walking is probably much less because the impact force during running is three to six times body weight (Hollerbach et al. 2000). Thus, the surface will deform much less than for running and hence affect the stance time much less. Our experiments involve only treadmill walking, albeit on the two different generations of Treadports, rather than treadmill walking versus ground walking. While it is possible that there is some small difference in surface compliance between the two Treadports, we expect any influence on our results to be minor.

With regard to underpowered treadmill belts, the dominant load to the belt motor is friction between the belt and its backing due to force from the user's foot. With impact forces of up to six times body weight during running and coefficients of friction around 0.15 between belt and backing, a substantial belt load force can be generated. The second-generation Sarcos Treadport's belt motor was designed so that a 90 kg user would not experience more than a 5% belt slowdown during running (Hollerbach et al. 2000). Treadmills that have not been adequately designed to maintain belt speed during footfall could conceivably perturb the biomechanics noticeably. Again, because our experiments involve walking and not running, any belt slowdown will be much less than 5%, and our experiments are only a within-treadmill study.

When a user accelerates on a treadmill rather than runs at constant velocity, there is a large energetic difference from ground running. Frishberg (1983) showed that runners use 35% less energy on a treadmill running a 100 yard dash. There are two likely causes of differences: viscous drag and lack of inertial forces. Since a user is stationary on a treadmill, there is a lack of viscous drag due to air resistance. The viscous drag is not large enough to explain this energetic difference and is not likely to be significant for walking.

Because the user's body is stationary with respect to the ground when accelerating on a treadmill, there is a missing inertial force that probably explains the energetic difference. Making a simple assumption that a human can be modeled as a point mass m, the missing inertial force is $f = ma$, where a is the user's acceleration. The tether can pull on the user based on measurements of belt acceleration to effectively supply this missing inertial force, thus making treadmill running energetically similar to ground running (Christensen et al. 2000). In the slope experiments, subjects walked at a constant velocity, and no inertial force feedback was used. Again, all experiments were performed on treadmills, so the effect of any small accelerations during walking would be similar.

There are other uses for tether force as well:

- Wall constraints can be displayed via penalty forces. This is very much like representing constraint surfaces with a haptic interface, but applied to the body rather than the hand. Merely stopping the treadmill belt when the user walks into an obstacle is not enough because the user would stumble forward.
- A spring-like centering force is applied to the user as a kinesthetic cue for safer operation. Noma and Miyasato (1998) define a locomotion interface as a motion canceling device, and an active mechanical tether can help by applying forces to recenter a user on the belt. The Omni-Directional Treadmill (ODT) employs a two-axis active mechanical tether for this purpose, which is ceiling mounted and can exert bias forces of up to 89 N.

The tether can also be used to simulate the extra gravity forces in slope walking by pulling or pushing on the user in the direction of walking.

1.3. Slope Display via Torso Force Feedback

When walking on a real slope, the gravity force f parallel to the slope that retards or assists walking is $f = mg\sin\theta$, where m is the user's mass, $g = 9.8m/s^2$ is gravity, and θ is the slope (see Fig. 4a). This gravity force can instead be applied by the mechanical tether to simulate slope walking (see Fig. 4b).

One difference between the two situations is the amount of force normal to the walking surface, which is $mg\cos\theta$ for normal slope walking but is always mg for walking on the level treadmill belt. Consequently, the total force magnitude f_t for level treadmill walking with horizontal tether force is as follows:

$$\begin{aligned} f_t &= \sqrt{(mg)^2 + (mg\sin\theta)^2} \\ &= mg\sqrt{1+\sin^2\theta}. \end{aligned} \tag{1}$$

The maximum slope in the experiments reported here is 14 degrees, in which case $f_t = 1.03mg$. The difference is fairly insignificant, especially in view of the experimental results that will show that subjects prefer a fractional slope force of $f = 0.65mg\sin\theta$. As speculated later, this fractional force preference may be due to the way forces are transmitted to the body by the tether and harness. Nevertheless, for steeper slopes, the higher resultant force could become a problem, and a vertical support force would be required. In a series of studies, Kram and colleagues employed a passive elastic vertical support system to simulate walking under reduced gravity conditions (Chang et al. 2000; Donelan and Kram 1997; Griffin, Tolani, and Kram 1999). We envision a future redesign of the active mechanical tether to provide not only vertical support forces but also sideways forces.

A related idea is to alter the pulling angle of the tether to be at an angle θ to mimic the effect of Figure 4a. As part of the redesign of the second-generation Treadport (Hollerbach et al. 2000), we tested subject discrimination of the pulling angle. Experiments reported in Tristano, Hollerbach, and Christensen (2000) showed a poor resolution of the tether pulling angle, on the order of 10 to 15 degrees. The inability of subjects to discern the pulling angle may be due to backlash in the harness. When subjects compared walking on a reference slope θ versus walking on a level slope but with tether force pulling angles of θ or zero degrees (horizontal), the horizontal pulling angle was generally preferred. Given these negative results on varying the tether force pulling angle, it was decided to keep the tether force horizontal in the redesigned Treadport of Figure 3. Not varying the tether pulling angle also greatly simplifies the tether design.

Because of its higher bandwidth, tether force can represent fast slope transients and is potentially a replacement for having a tilt mechanism at all. The absence of a tilt mechanism would have the added advantage of simplifying video displays that use configurations such as back-projected screens. As seen in Figure 3, there is a three-wall CAVE-like (Cruz-Neira, Sandin, and DeFanti 1993) visual display in a flared arrangement spanning a 180-degree field of view in front of the Treadport. If the platform tilts up and down, then portions of the screen are covered or uncovered. The belt of the Treadport is white to allow for the possibility of floor projection, but tilting would distort the image and require computation to predistort the image depending on the tilt angle. Although the use of a head-mounted display (HMD) would obviate these problems, we do not prefer HMDs for the Treadport because

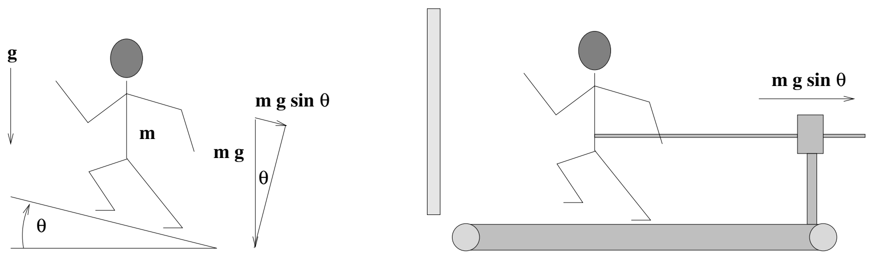

Fig. 4. (a) Gravity force $f = mg \sin\theta$ opposes uphill walking. (b) Simulation of this gravity force with an active mechanical tether.

of inferior field of view, safety, and comfort and the inability to see one's body as part of the virtual world.

The purpose of this paper is to quantify how well in fact the tether force can simulate slope walking, using psychophysical and biomechanical experiments. There is not a large preexisting literature on human slope walking, and certainly nothing so far on the use of torso force feedback for that purpose.

To begin, we are not aware of any experiments that have been done to quantify how sensitive humans are to slope inclination, perhaps because there was no reason to ask the question before. There is a large literature on the perception of joint angles (Clark and Horch 1986), although not in terms of a task such as walking. The implication of a knowledge of slope sensitivity while walking is a design specification for representing slope on a locomotion interface. The first experiment that we report on quantifies through a discrimination task how sensitively people judge slope changes while walking on the treadmill.

Next, we perform psychophysical experiments on slope walking versus tether force walking to judge the equivalence of the experience. In these experiments, subjects walk on a particular slope θ, then walk on a level slope with tether force f adjusted according to their perception of the best "equivalence" of the tether force level. The issue is whether the predicted relation $f = mg \sin\theta$ is satisfied.

A more quantitative test would be to show biomechanical equivalence (i.e., the gait patterns are the same for the two situations). It is not unreasonable to expect a biomechanical correlate because a user has to lean against the tether force in a manner that could conceivably be similar to leaning while walking on a slope. This paper presents such a biomechanical analysis. Past research on the biomechanics of slope walking have employed various kinematic measures to quantify the change of gait with slope, such as leg joint angle ranges (Masmoudi et al. 1999) and the knee-hip cyclogram (Goswami 1998). We have examined a range of measures to deduce what is the best biomechanical correlate for slope and then used such a measure to examine whether the biomechanics of slope walking versus tether force walking are similar. Portions of this research have been previously reported (Tristano, Hollerbach, and Christensen 2000; Mills, Hollerbach, and Thompson 2001).

2. Discrimination of Slope during Walking

To set a baseline for the interpretation of results with tether force simulation of slope, we wanted to know how well humans can discriminate slope differences while walking. Interestingly, the literature did not have this information, perhaps because there was no reason to ask this question before. The following experiment was devised to answer this question, employing the original Treadport (see Fig. 1).

Subjects were asked to walk on the treadmill at a constant pace at different slopes. From a particular reference slope, the slope of the treadmill was adjusted either up or down, and subjects were asked to indicate the direction of the change. When the slope was changed, the treadmill belt was stopped and the subjects remained standing on the treadmill. The order of presentation of slopes was randomized. Subjects wore headphones to help mask the sound of the tilt motor, although it appeared that the motor made the same noise whether the platform went up or down. Preliminary tests were done in which the up-down motion was randomized before settling at the final slope to avoid kinesthetic cues about the adjustment. This random excursion stayed within 0.5 degrees and lasted approximately 10 to 12 seconds. The results were similar to just placing the treadmill at the final slope, which takes 2 seconds, and for the results reported below the treadmill motions were not randomized.

Subjects kept their eyes closed. To provide kinesthetic cues to help them stay centered while their eyes were closed, a string attached to two posts at the front of the treadmill was held by the subjects (see Fig. 5). The string attachment points at the posts were through springs, to approximate a constant force source when pulling on the string. This avoids the use of any force cues in pulling on the string in judging slope. The spring constants were low, so that the subject could not pull as an aid in walking up slopes.

Five different reference slopes were used: 0 degrees, ±2 degrees, and ±4 degrees. For each reference slope, there were

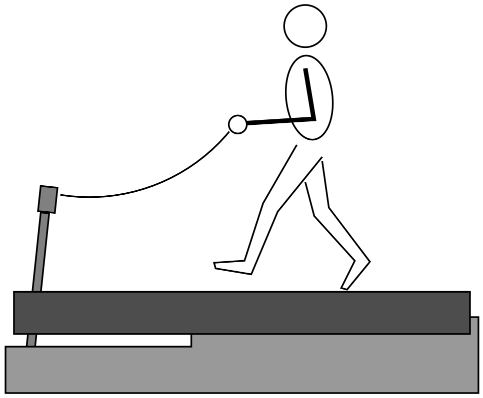

Fig. 5. A string guide for visionless walking.

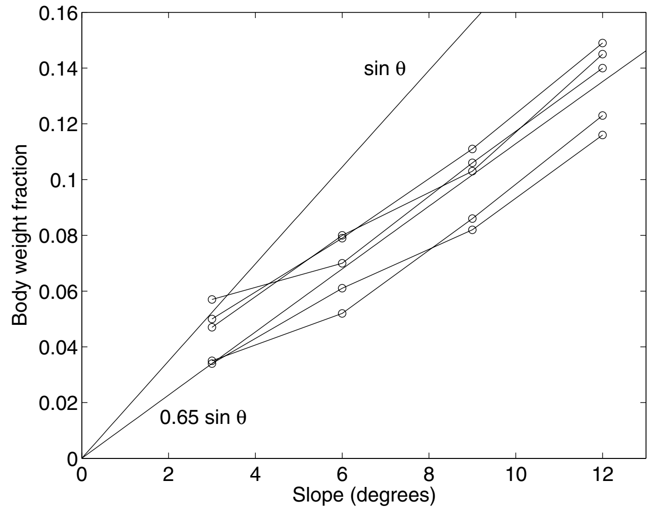

Fig. 6. Reference slope versus tether force expressed as a fraction of body weight. Slope increments are 3 degrees from 3 to 12. Data are from 5 subjects. The upper straight line is a plot of $\sin\theta$, where θ is the slope in radians. The lower straight line is a plot of $0.65 \sin\theta$.

six different adjustments: ± 0.5 degrees, ± 0.75 degrees, and ± 1.0 degrees. There was not an option of the slope remaining the same, and subjects made a forced binary choice of higher or lower. The order of the trials was randomized and then fixed for all subjects. There were 13 subjects (10 male and 3 female) ranging in age from 20 to 50. Subjects were given as much time as they wished. No knowledge of results was provided. The entire experiment lasted approximately 15 minutes.

Table 1 shows the percentage of correct responses for all subjects for slope discrimination, combining all reference slopes. The results are not strictly correlated with the magnitude of slope change. With a forced binary-choice paradigm, strict guessing would yield a 50% success rate, and so traditionally a criterion of 75% is considered as the threshold of discrimination. Consequently, given that the success rates ranged from 77% to 89%, subjects were able to perceive a 0.5 degree slope change.

Table 2 shows the success rate as a function of the reference slope for all subjects, combining all adjustments. People were most accurate when starting from level. There is not a substantial difference between the 2- and 4-degree reference slopes.

There was evidence that learning occurred. The success rate for the first 15 trials was 77%, while the rate for the last 15 trials was 87%. Subjects by and large felt they were guessing the whole time and were surprised as to the extent of their correct responses.

The sensitivity of 0.5 degrees of subjects to slope changes is remarkable in its precision. Certainly, some form of accurate slope presentation is necessary for locomotion interfaces.

3. Tether Force versus Perceived Slope

The next experiment investigated psychological equivalence between real slope walking versus walking on a level slope but being pulled by the tether. The original Treadport (see Fig. 1) was again employed. Subjects walked on a reference slope for a minute, and then the slope was returned to horizontal. The pulling force was varied according to a subject's directive, until the subject had found that the magnitude of pulling force was most appropriate to duplicate the sensation of the slope. Slopes were chosen in 3-degree increments, from 3 to 12 degrees upwards. Experiments lasted about half an hour.

The results for 5 subjects are plotted in Figure 6 as reference slope (horizontal axis) versus force represented as a fraction of body weight. According to Figure 4, we would predict a hypothetical relation $f/mg = \sin\theta$, where θ is the reference slope represented in radians and f/mg is the fraction of body weight. This hypothetical relation is plotted as the upper "straight" line in the figure; of course, the line is not exactly straight, but for small angles $\sin\theta \approx \theta$. It is seen that the slopes of the experimental plots are less than the hypothetical relation.

A least squares fit of f/mg versus $\sin\theta$ was performed, with the constraint that the fitted line pass through the origin. The result is that the slope $s = 0.65$, and hence $f/mg = 0.65 \sin\theta$ (i.e., approximately two-thirds of the expected force). The standard deviation σ of the straight-line fit is 0.0128, and the estimate σ_s of the standard deviation of the slope s is 0.02. If all of the data are placed into single columns Y for f/mg and X for $\sin\theta$, then $\sigma^2 = (Y - sX)^2/(N-1)$, where N is the number of data points. The estimate σ_s is computed from the standard relation $\sigma_s^2 = \sigma^2(X^T X)^{-1}$ (Hollerbach and Wampler 1996). The standard deviation of the estimated slope $\sigma_m = 0.02$ is fairly small compared to the estimate $s = 0.65$, which indicates a reasonably good fit. Hence,

Table 1. Success Rate for All Subjects for Slope Discrimination

Angle change (deg)	−1.0	−0.75	−0.5	0.5	0.75	1.0
Success rate (%)	85	83	77	82	89	77

Table 2. Success Rate versus Reference Slope

Reference angle (deg)	−4	−2	0	2	4
Success rate (%)	73	83	94	77	85

tether force is a reasonable means for slope display based on psychological measures.

However, the preferred proportion of tether force is two-thirds of the hypothetical. One explanation is that gravity loads and inertial loads should be distributed over the whole body, but the tether just applies force to one point at the back of the body and through the harness. Another explanation is that the segmental mechanics of the body are not being modeled; instead, the body is simply lumped into an undifferentiated mass m. The next study shows that the two-thirds value is highly relevant and not just some random feature from a psychological study.

4. Biomechanics of Slope versus Tether Force Walking

While psychological equivalence is one way to demonstrate that tether force reasonably simulates slope, a more objective measure would be preferable. The question is whether we walk the same way under the two circumstances, which was quantified by biomechanical measures of gait.

Measurement of gait was done with the Northern Digital Optotrak System, which involves placement of active LED markers on the foot, calf, thigh, and hip (see Fig. 7). Special rigid bars for LED mounting were created to facilitate joint angle calculation by considering these bars as vectors representing absolute orientation of leg segments. Padding and straps were employed to ensure tight but comfortable coupling to the limbs.

Two different generations of Treadport were employed in this study. The second-generation Treadport has a redesigned belt drive and mechanical tether, which are improvements over the first-generation Treadport (Hollerbach et al. 2000), but does not yet have a functioning tilt mechanism. Therefore, we employed the first-generation Treadport (see Fig. 1) to generate a tilted walking surface and the second-generation Treadport to apply tether forces (see Fig. 3).

The gaits of 6 subjects (3 male and 3 female) were measured while walking on a tilted belt and while walking on a flat belt but with tether force application. The age range was

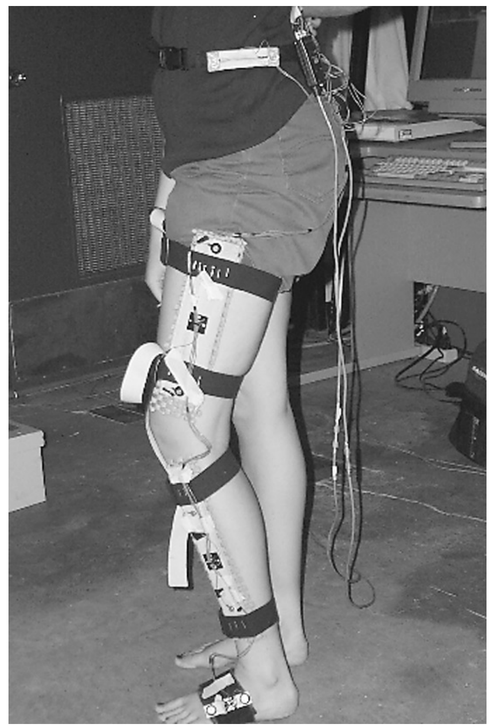

Fig. 7. Marker attachment for leg joint angle measurements.

21 to 54. The tether force was applied to a user via a torso harness to which the active mechanical tether attaches. The tether uses a linear drive consisting of a timing belt and a geared electric motor (Hollerbach et al. 2000) and is capable of exerting 315 N.

For the treadmill tilt experiments, the subject walked on the first-generation Treadport, and the slope of the Treadport was varied randomly at 2-degree intervals between 6 degrees downhill and 14 degrees uphill. This range was dictated by the asymmetry of the tilting mechanism of the first-generation Treadport. For the tether force experiments, the subject walked on the second-generation Treadport, and forces on the tether were varied randomly from −100 N to 45 N. A negative force corresponds to a force pulling the subject and therefore simulates a positive slope, and a positive force simulates a negative slope accordingly. Both Treadports were kept at a constant walking speed throughout the experiment. Four trials were run for each slope or tether force value, and each trial lasted 4 seconds, which was enough for two to three cyclograms.

The data collected are similar across all subjects. Hence, we present representative results of one specific subject to show the trends and characteristics that are common to all subjects in the experiment. Knee and hip angles were derived from the positions of the sensors and then plotted against one another. These plots are the knee-hip cyclograms found in Figures 8 and 9. Figure 8 shows how cyclograms change according to variation in slope. As the slope increases, the cusp of the cyclogram rotates clockwise, the knee and hip angle ranges widen, and the overall shape of the cyclogram becomes more oblong. The cusp happens at footfall, where the knee flexes almost elastically before straightening out to push off.

Figure 9 shows how the cyclograms change according to various tether forces applied to the subject. The cyclogram trends are similar to those of Figure 8; the cusp rotates clockwise, knee and hip angle ranges widen, and the overall shape of the cyclogram becomes more oblong.

These visual changes and trends can be captured quantitatively by feature analysis of the cyclograms (Goswami 1998) and by joint angle ranges. Using this analysis, the following statistics were calculated: hip range, knee range, ratio of hip range/knee range, ratio of knee range/hip range, area of cyclogram, circularity, eccentricity, orientation, and cusp orientation. These values are shown in Figures 10 and 11. We reiterate that negative tether force simulates a positive treadmill slope, and so the slopes of Figure 11 are reflections of those in Figure 10. Once again, the general trends of the cyclogram are similar as the slope and tether force change from smaller slopes to larger slopes.

To find a relationship between tether force and the simulated slope angle, least squares equations were found for each of the properties of the cyclogram for all subjects. The most linear properties across both slope and force were the hip range, the knee/hip range ratio, the cyclogram orientation, and the cyclogram cusp orientation. Analysis of variance accounted for showed that hip range is consistently the most linear feature with slope or force and hence is used in the subsequent analysis. Straight-line fits were made to hip range versus force, $HR = af + b$, and hip range versus slope, $HR = c\theta + d$, for each subject. The approximation $\theta \approx \sin\theta$ was used, which only has a 1% error at the maximum tilt of 14 degrees. Then a relation between the tether force and slope could be predicted as follows:

$$f = \frac{c}{a}\theta + \frac{d-b}{a}. \qquad (2)$$

Table 3 shows the results for the 6 subjects. We are expecting a relation $f = mg\sin\theta$, so the intercept $(d-b)/a$ not being zero is an indication of the approximation of the linear fits.

Table 3. The Slope *c/a* and Intercept (*d* – *b*)/*a* of the Experimentally Derived Linear Relation between Tether Force and Slope and the Fractional Force Result *c/a/mg*

Subject	c/a	(d – b)/a	c/a/mg
1	−437	2.0	0.647
2	−289	−15.5	0.520
3	−494	−19.8	0.756
4	−444	−17.9	0.526
5	−576	32.0	0.730
6	−480	3.6	0.663

As mentioned earlier, it was found from psychophysical experiments that there was a fractional force preference of 65% of the full predicted gravity force (i.e., $f = 0.65mg\sin\theta$). By dividing the slope c/a by a subject's weight mg in eq. (2), we will be able to tell whether the biomechanical results predict a partial force as well. The last column in Table 3 shows these fractional forces, which range from 52% to 73% across subjects. These results are in the vicinity of the average result of 65% found from the psychophysical studies.

Consequently, a fractional force of roughly 65% must be the correct force level to apply to the subjects since it has now been verified and determined from an objective biomechanical measurement. This force level is reflecting how forces are actually being applied to the body to simulate the gravity force in slope walking.

5. Discussion

The main result of this paper is to demonstrate that tether force is a reasonable means for displaying gravity force, which has been shown both through psychological equivalence and biomechanical equivalence. In the psychological experiments, the fit of tether force to equivalent treadmill tilt approximately satisfies the relation $f = 0.65mg\sin\theta$, or two-thirds of the hypothetical. The biomechanical experiments demonstrate that the horizontal tether force changes a person's gait in a manner that is similar to the gait changes of the person walking on different slopes.

Several gait features were found to have approximately linear relationships with slope or tether force: hip range, the knee/hip range ratio, the knee/hip cyclogram orientation, and the knee/hip cyclogram cusp orientation. Of these, the hip range was the more linear. Hip range was also noted as an important slope indicator by Masmoudi et al. (1999), although the plots of hip range versus slope were not as linear as what we found. Goswami (1998) had previously characterized higher order moments of the knee/hip cyclograms as

(text continues on p. 951)

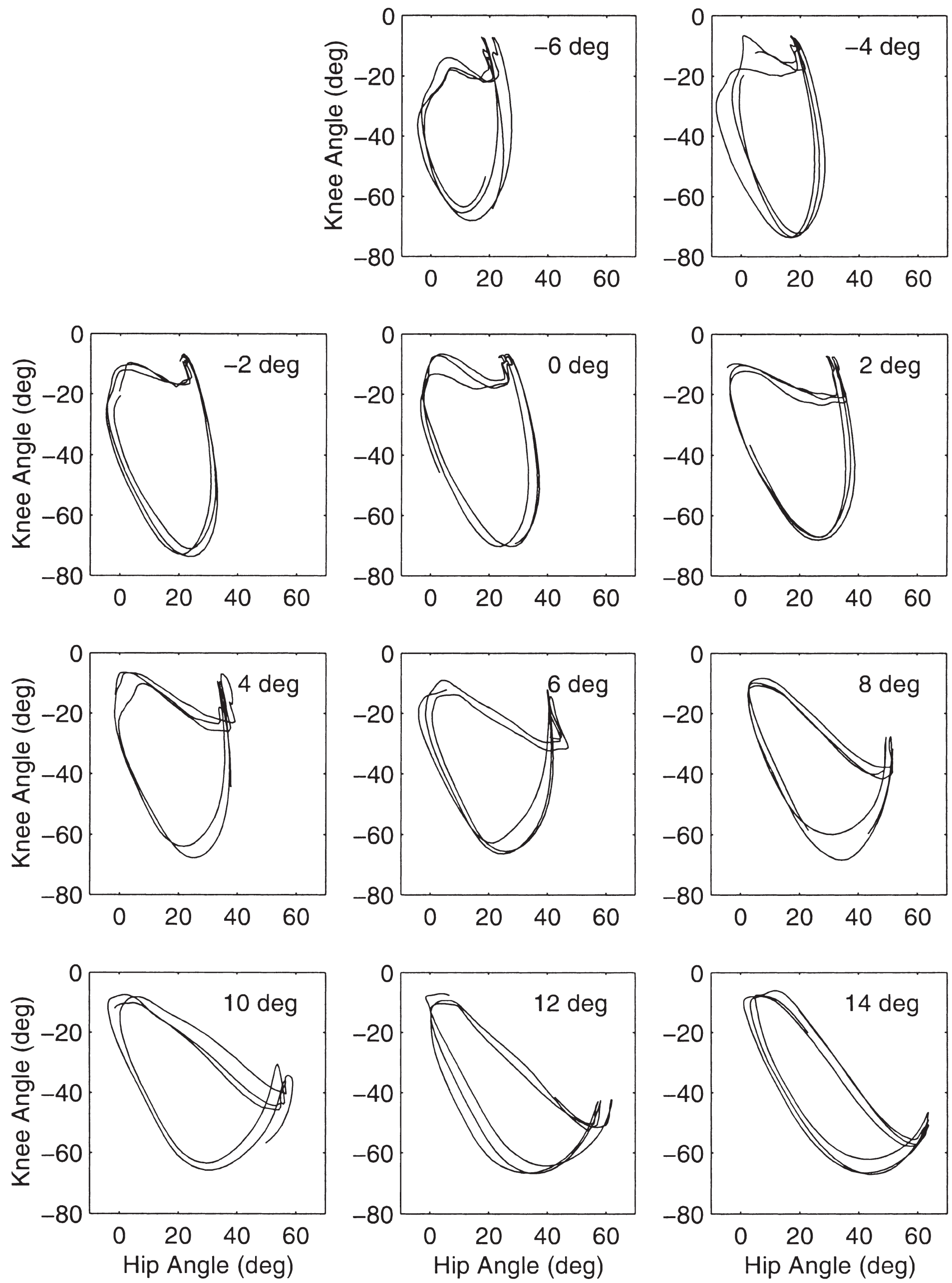

Fig. 8. Hip-knee cyclograms at different slopes.

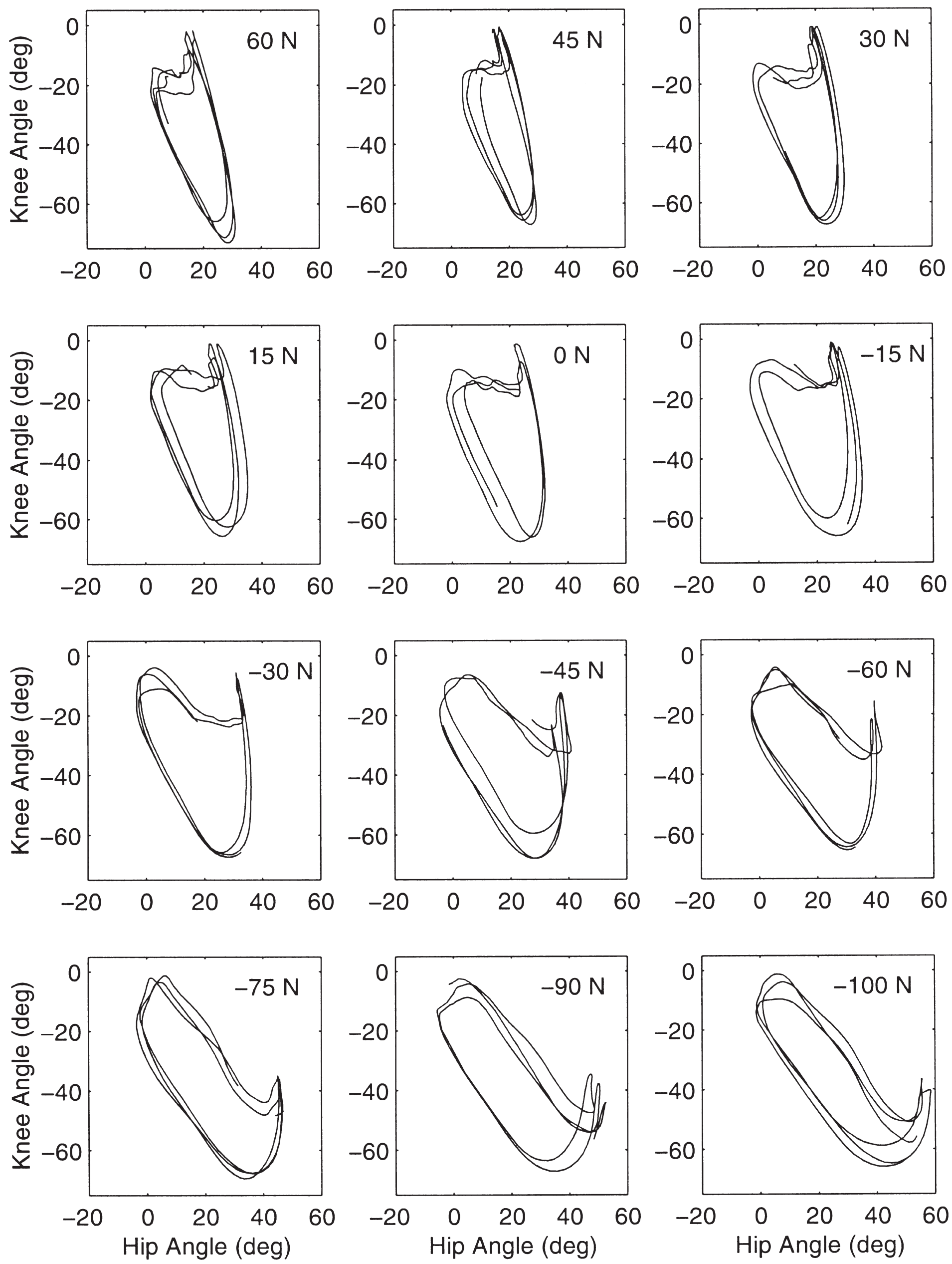

Fig. 9. Hip-knee cyclograms at different tether forces.

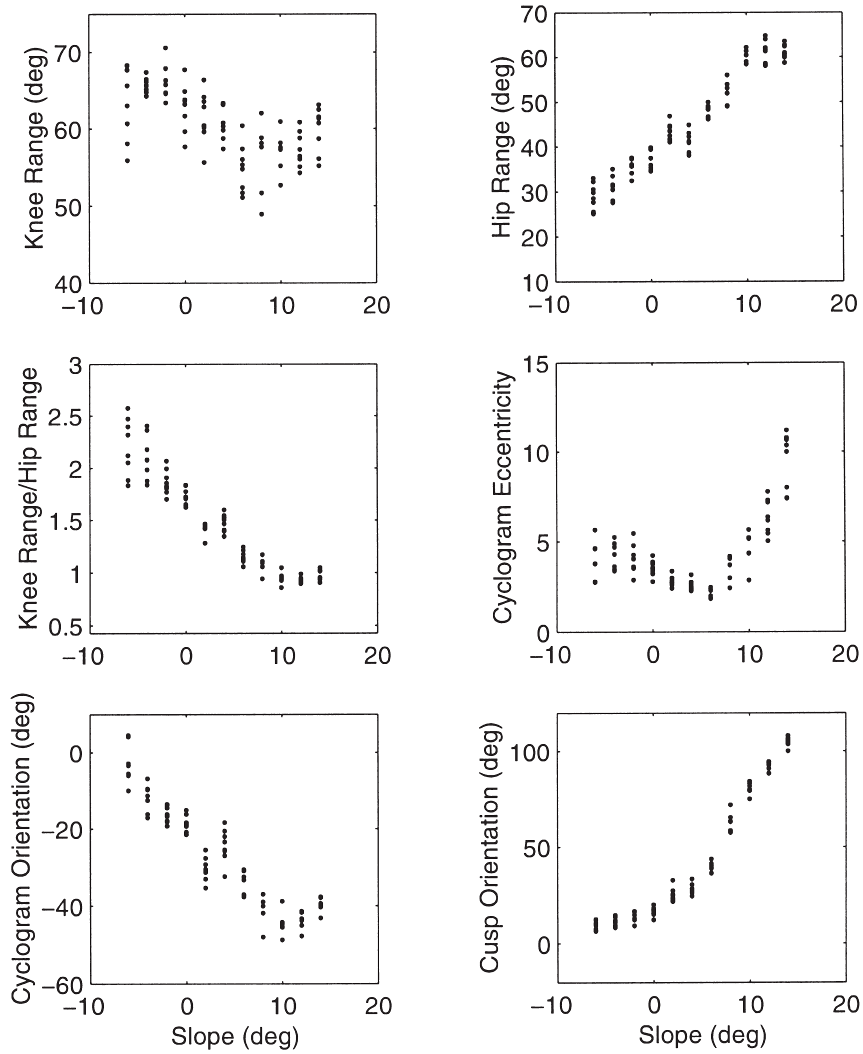

Fig. 10. Properties of cyclograms as they change according to slope.

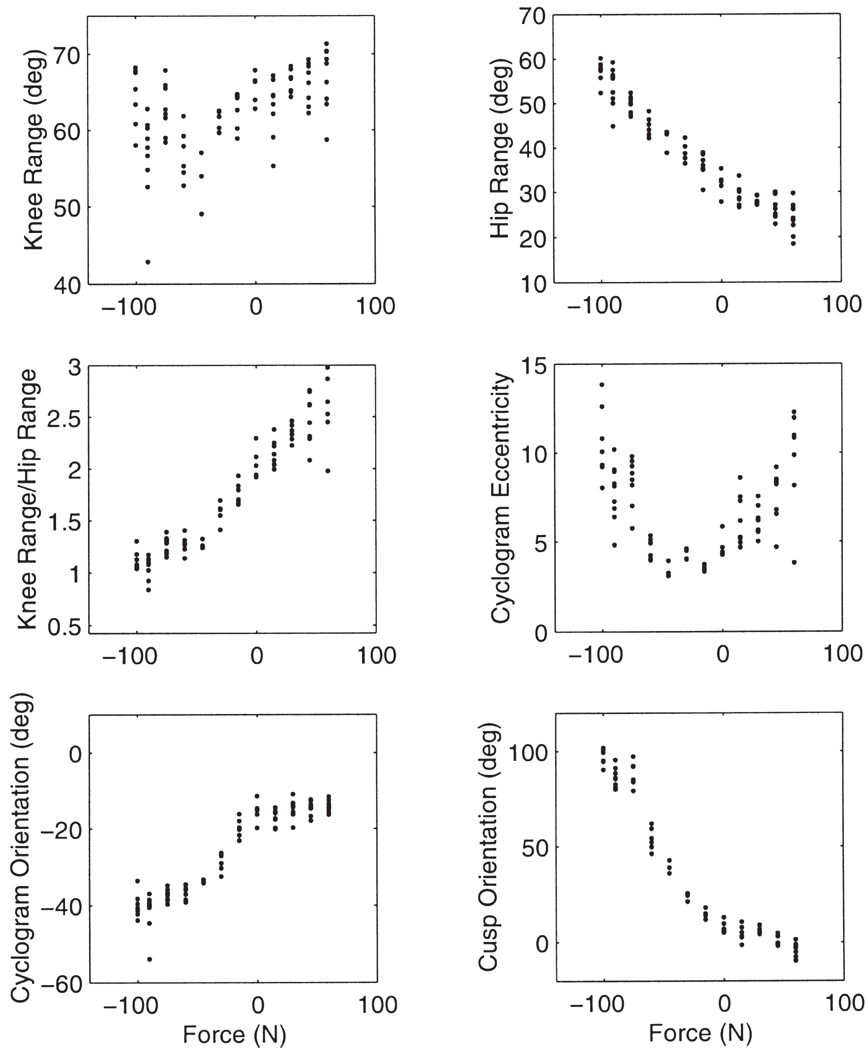

Fig. 11. Properties of cyclograms as they change according to tether force.

good descriptors of slope walking. Our work has shown that the orientation of the knee/hip cyclogram as a whole and the orientation of just the cusp part of the knee/hip cyclogram provided good linear characterizations of slope walking.

The close linear fits of tether force to hip range and treadmill tilt to hip range allowed a prediction of tether force to treadmill tilt. By dividing the slope of the linear relation of tether force to treadmill tilt angle by each subject's weight, it was found that a fractional application of force between 52% and 73% of the expected amplitude $f = mg \sin\theta$ was appropriate to represent a particular slope θ. This result is consistent with the 65% fractional preference determined from psychophysical experiments. Because the fractional forces were derived from biomechanics, it must indeed be the case that the proper tether force is not 100% of the predicted gravity force $f = mg \sin\theta$.

As mentioned earlier, the cause of the fractional force must have something to do with the point-force application to the body by the tether or by the method of force distribution to the body by the harness. An analogous result was observed in subjects' preference for inertial force display (Christensen et al. 2000). The hypothetical inertial force feedback was $f = ma$, where a was the subject's acceleration, but the preferred relation was $f = 0.8ma$ (i.e., 80% of the hypothetical). Actually, the 80% figure was not determined so precisely as the 65% figure, and so an exact comparison is not yet appropriate. An explanation for the fractional force based on a mechanical analysis awaits future analysis.

There is an implication for treadmill design because one can now choose between treadmill tilt and tether force to simulate slope. As mentioned earlier, there are already reasons to include an active mechanical tether with treadmills when creating locomotion interfaces, such as inertial force display (Christensen et al. 2000), the display of hitting objects, safety restrictions to a range of forward motion on the treadmill surface, and accurate tracking of user position. One can then add to that list the accurate display of slope.

Although treadmill tilt simulates slope realistically, there are some reasons against tilt implementations. The treadmill surfaces are large and heavy, especially the large 6-by-10 foot surface of the Sarcos Treadport II, and so the tilt mechanism adds cost and complexity to the design and will be slower than the fast-acting mechanical tether. When using a stationary CAVE visual display, a tilted platform would obscure portions of the screens. An alternative is to mount the CAVE on the treadmill (H. Noma, personal communication, 2001), although the size of the display will be necessarily limited. If projection onto the belt surface is contemplated, then the image will be distorted and will have to be compensated for by computation.

The results on slope perception of 0.5 degrees while walking also point to a high sensitivity to slope while walking. While tether force can substitute for treadmill tilt, it may still be desirable also to have a fast-tilt mechanism. It is possible that tilt and tether force can be combined and blended to display sudden slope changes that then persist.

An application of these results besides virtual reality is in the use of treadmills for legged robot research. For example, treadmills have been built for the running robots of Raibert (1986) and Moghaddam and Buehler (1993). Our results suggest that slopes could be simulated for the running robots by adding an active mechanical tether. In addition, the mechanical tether could supply realistic inertial forces to those robots (Christensen et al. 2000).

This research could be extended in several ways. Besides walking, the biomechanical characteristics of standing and running on a slope versus with tether force could be compared to confirm the equivalence. If a sideways tether force could be applied, it would be of interest to check if side-slope walking could also be simulated with tether force. We only tested constant slopes, but there is a question as to whether time-varying tether forces could represent uneven slopes, perhaps even stairs. Finally, as mentioned in the introductory section, if a vertical tether force could be applied, then very steep slope walking could be more realistically simulated by correct resultant force application.

Acknowledgments

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References

Chang, Y. H., Huang, H.W.C., Hamerski, C. M., and Kram, R. 2000. The independent effects of gravity and inertia on running mechanics. *Journal of Experimental Biology* 203:229–238.

Christensen, R. R., Hollerbach, J. M., Xu, Y., and Meek, S. G. 2000. Inertial-force feedback for the Treadport locomotion interface. *Presence: Teleoperators and Virtual Environments* 9:1–14.

Clark, F. J., and Horch, K. W. 1986. Kinesthesia. In *Handbook of Perception and Human Performance: I. Sensory Processes and Perception*, ed. K. R. Boff, L. Kaufman, and J. P. Thomas, 13.1–13.62. New York: John Wiley.

Cruz-Neira, C., Sandin, D., and DeFanti, T. A. 1993. Surround-screen projection-based virtual reality: The design and implementation of the CAVE. *ACM SIGGRAPH'93 Proceedings*, Anaheim, CA, August, pp. 135–142.

Dal Monte, A., Fucci, S., and Manoni, A. 1973. The treadmill used as a training and simulator instrument in middle- and long-distance running. *Biomechanics III: Medicine and Sport* 8:359–363.

Darken, R. P., and Cockayne, W. R. 1997. The Omni-Directional Treadmill: A locomotion device for virtual worlds. *Proceedings of UIST*, pp. 213–221.

Donelan, J. M., and Kram, R. 1997. The effects of size, speed and gravity on the kinematics of human walking: A test of the dynamic similarity hypothesis. *Journal of Experimental Biology* 200:3193–3201.

Durlach, N. I., and Mavor, A. S., eds. 1994. *Virtual Reality: Scientific and Technological Challenges*. Washington, DC: National Academy Press.

Elliot, B. C., and Blanskby, B. A. 1976. A cinematograhpical analysis of overground and treadmill running by males and females. *Medicine and Science in Sports and Exercise* 8:84–87.

Frishberg, B. A. 1983. An analysis of overground and treadmill sprinting. *Medicine and Science in Sports and Exercise* 15:478–485.

Goswami, A. 1998. A new gait parameterization technique by means of cyclogram moments: Application to human slope walking. *Gait & Posture* 8:15–36.

Griffin, T. M., Tolani, N. A., and Kram, R. 1999. Walking in simulated reduced gravity: Mechanical energy fluctuations and exchange. *Journal of Applied Physiology* 86:383–390.

Hirose, M., Ogi, T., Yano, H., and Kakehi, N. 1999. Development of wearable force display (HapticGEAR) for immersive projection displays. *Proceedings of IEEE Virtual Reality*, Houston, TX, March, p. 79.

Hollerbach, J. M. 2002. Locomotion interfaces. In *Handbook of Virtual Environments Technology*, ed. K. Stanney, 239–254. Englewood Cliffs, NJ: Lawrence Erlbaum.

Hollerbach, J. M., and Wampler, C. W. 1996. The calibration index and taxonomy for robot kinematic calibration methods. *International Journal of Robotics Research* 14:573–591.

Hollerbach, J. M., Xu, Y., Christensen, R., and Jacobsen, S. C. 2000. Design specifications for the second generation Sarcos Treadport locomotion interface. *Haptics Symposium, Proceedings of ASME Dynamic Systems and Control Division*, DSC-Vol. 69-2, Orlando, FL, November, pp. 1293–1298.

Ingen Schenau, G. J. Van. 1980. Some fundamental aspects of the biomechanics of overground versus treadmill locomotion. *Medicine and Science in Sports and Exercise* 12:256–261.

Iwata, H. 2000a. Locomotion interface for virtual environments. In *Robotics Research: The Ninth International Symposium*, ed. J. Hollerbach and D. Koditschek, 275–282. London: Springer-Verlag.

Iwata, H. 2000b. The haptic interfaces of the next decade. *Proceedings of IEEE Virtual Reality 2000*, New Brunswick, NJ, March, pp. 241–243.

Iwata, H., and Yoshida, Y. 1999. Path reproduction tests using a Torus Treadmill. *Presence* 8:587–597.

Masmoudi, L.-A., Martin, L., Cordier, E., and Blanchi, J.-P. 1999. Kinematics analysis of human slope walking. *First International Conference on Science & Technology in Climbing and Mountaineering*, Leeds, UK.

McMahon, T. A. 1984. *Muscles, Reflexes, and Locomotion*. Princeton, NJ: Princeton University Press.

Mills, R., Hollerbach, J. M., and Thompson, W. B. 2001. The biomechanical fidelity of slope simulation on the Sarcos Treadport using whole-body force feedback. In *Experimental Robotics VII*, ed. D. Rus and S. Singh, 437–446. London: Springer-Verlag.

Moghaddam, M., and Buehler, M. 1993. Control of virtual motion systems. *Proceedings of IEEE/RSJ International Conference on Intelligent Robots and Systems*, Yokohama, Japan, July, pp. 63–67.

Nelson, R. C., Dillman, C. J., Lagasse, P., and Bickett, P. 1972. Biomechanics of overground versus treadmill running. *Medicine and Science in Sports* 4:233–240.

Nigg, B. M., Deboer, R. W., and Fisher, V. 1995. A kinematic comparison of overground and treadmill running. *Medicine and Science in Sports and Exercise* 27:98–105.

Noma, H., and Miyasato, T. 1998. Design for locomotion interface in a large scale virtual environment. ATLAS: ATR Locomotion Interface for Active Self Motion. *Proceedings of ASME Dynamic Systems and Control Division*, DSC-Vol. 64, pp. 111–118.

Noma, H., Sugihara, T., and Miyasato, T. 2000. Development of ground surface simulator for tel-E-merge system. *Proceedings of IEEE Virtual Reality 2000*, New Brunswick, NJ, March, pp. 217–224.

Raibert, M. H. 1986. *Legged Robots That Balance*. Cambridge, MA: MIT Press.

Sheridan, T. B. 1992. *Telerobotics, Automation, and Human Supervisory Control*. Cambridge, MA: MIT Press.

Tristano, D., Hollerbach, J. M., and Christensen, R. 2000. Slope display on a locomotion interface. In *Experimental Robotics VI*, ed. P. Corke and J. Trevelyan, 193–201. London: Springer-Verlag.