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Implementing Virtual Stairs on Treadmills Using Torso Force Feedback

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Abstract—**This paper describes the simulation of stairs on a treadmill style locomotion interface using torso force feedback. The active mechanical tether of the Sarcos Treadport locomotion interface applies a specialized force profile to simulate the forces of stair walking. The biomechanics of subjects walking on real stairs versus walking under the specialized force profile were compared. It was found that the tether force was able to adjust the subject's motion from standard slope walking towards that of stairs.**

I. INTRODUCTION

This paper discusses a method for simulating stairs on a treadmill style locomotion interface using torso force feedback. The Treadport is an example of an active treadmill locomotion interface [7], where measurements of user position and orientation control the treadmill speed and walking direction. The Treadport also employs an active mechanical tether which attaches to the small of the user's back through an harness as shown in Figure 1. The tether can apply a force along its linear axis either forward or backwards, and has previously been used in our research to simulate inertial forces [3] and slope forces [16], [12], and to provide constraint forces such as walls. Torso force feedback has been used in other locomotion interfaces such as the Omni-Directional Treadmill to provide a centering force on the platform [4] and by Kram using passive rubber tubes to generate horizontal forces [1] and vertical forces [2]. In this paper we introduce a new application for torso force feedback, simulating stairs.

Other non-treadmill style locomotion interfaces have shown success in simulating stairs by using mechanical foot platforms. The Sarcos Biport employs two three degree of freedom hydraulically actuated arms on which a user stands to simulate rigid surfaces such as stairs [7]. The GaitMaster, another foot platform device developed by Iwata [8], was able to accurately simulate stairs at a slow pace. However, in both these cases the device was only able to simulate stairs, but not slopes or uneven terrain because the foot platforms were fixed parallel to the floor.

Another approach for displaying steps on a linear treadmill is ATR's Ground Surface Simulator (GSS), which employs a flexible belt that is deformed underneath by six vertical stages [13]. The belt is 1.5 m long and 0.6 m wide. Each stage is 0.25 m long and has a stroke of 6 cm at a speed of 6 cm/s. There are rollers on the support surface of each stage. Because the geometry of the belt changes when deformed by the stages, an active belt tensioning system is employed. A slope of 5 degrees can be presented by the GSS.

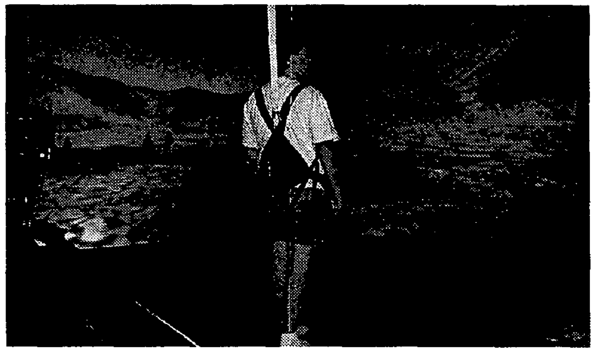

Fig. 1. The Sarcos Treadport locomotion interface is based on a linear treadmill whose belt is 10 feet long and 6 feet wide. The treadmill is augmented with a mechanical tether that measures user position and exerts a horizontal force in the forwards-backwards direction. The primary visual display is a 3-wall back-projected display system in a flared arrangement. Shown is a simulation of walking in the mountains near Snowbird, Utah. A ceiling strap acts as a safety restraint system.

Our own research has focused on developing the capabilities of linear treadmill style locomotion interfaces, as represented by the Sarcos Treadport, and in particular on the use of torso force feedback. Linear treadmills allow realistic and relatively unfettered forward motion [3]. Their main limitations are the control of turning, and absent the belt deformation approach of the GSS the display of uneven terrain. The limitations of turning control have been addressed to some extent [17]. As mentioned earlier, it has been shown that the active mechanical tether of the Treadport can realistically simulate smooth slope [16], [12]. Uneven terrain is characterized by rapidly changing slope, and it is not unreasonable to expect that a time-varying tether force profile could simulate uneven terrain walking. One common form of uneven terrain is stairs, and this paper considers how well stairs can be simulated using just horizontal tether force. Ideally a vertical force capability would also be employed, but the Treadport does not yet have this capability.

II. Stairs vs Virtual Stairs

For level walking the gait cycle starts with the heel strike of one foot and ends with the next heel strike of the same foot. For stairs the heel or toe could possibly strike first so the gait is defined as first foot strike (whether heel or toe) and the next first strike of the same foot. The gait cycle is furthermore broken up into two phases, a swing phase and stance phase. Studies have shown that for level walking the stance phase is roughly 60% of the gate cycle, while for ascending stairs it is 66% [9], [15]. Livingston [10], however, showed that the stance and swing phase proportions varied significantly with the dimensions of the stair steps and the subject's height.

A subject's general biomechanical motion is also affected by these factors, thus an intra-subject comparison of biomechanics is appropriate. The biomechanical variables considered are the individual leg segment orientations and body angles, and these variables are processed in terms of knee range, hip-knee cyclograms [5], and upper body orientation. Our earlier research employed the hip angle range to equate a constant force slope simulation to the true slope to show the effectiveness of slope simulation using tether force [12]. Knee flexion is considerably greater in stair climbing motion than in level walking (12 degrees from one study [6]) with the range of motion again depending on the stair height and subject height [10].

In previous research on slope simulation using torso force feedback, the human was modeled as a cart. The component of the user's weight parallel to a slope of angle ϕ is $mg\sin(\phi)$, where m is the user's weight and g is the gravity constant. This is the force applied by the tether in a horizontal direction to simulate slope walking. Biomechanical and psychophysical experiments showed that a fractional force application of 65% of the full force $mg\sin\phi$ is actually the correct force magnitude [12], [16]. A possible reason for the reduced force amount is the concentration of the force at one point on the back instead of on the whole body.

The cart model, however, is too simplistic to simulate stair walking. Instead, we modify a spoked wheel model, which has been shown to be an effective approximation for bipedal walking [11]. The human is modeled as a point mass moving on a variable length pivot arm, which represents the pivot leg (Figure 2). The stair height is sh, the height of the user's center of gravity above the previous stair is uh, and the horizontal distance from the user's center of gravity to the foot placement is w. Assuming regular stairs and walking pattern, the angle of the leg with respect to the horizontal surface of the stair is $\theta = \arctan((uh - sh)/w)$.

Similar to before, a person's weight is split into two components, one into the stair along the direction of the pivot leg $f_{foot} = mg\sin(\theta)$ and the other perpendicular to the pivot leg in the direction back down the stairs

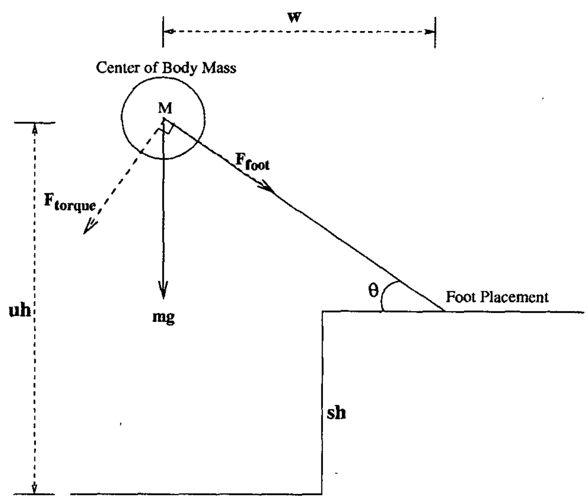

Fig. 2. Stair force model. The step height is sh, the user height above the previous stair is uh, and the horizontal distance from the user's center of gravity to the foot placement is w. The extended leg makes an angle θ with the next stair. The component of weight along the extended leg is $f_{foot} = mg\sin\theta$, while the component parallel to the stair direction is $f_{torque} = mg\cos\theta$.

$f_{torque} = mg\cos(\theta)$. f_{foot} is the force the leg must support while climbing up to the next step, while f_{torque} is the force that a human must overcome to propel up to the next step. This model is employed to predict f_{torque} during one step up the stairs, where the tether will apply f_{torque}. The next section details the experiments performed to calculate this force profile using each subject's motion on real stairs to tailor the forces applied for each individual.

When applying the variable stair force it is essential to have a strong mechanical coupling between the body and the tether, otherwise the effect of the force applied will be lessened as the force is damped. Changes were made to the harness to improve the mechanical coupling by moving and adding straps to cinch around the hips and chest.

The last necessary feature for simulating stairs is to apply the calculated force profile at the correct time. To accomplish this we needed to track the subject's feet to determine when the foot hits the ground (corresponding to the foot landing on the next step in real stairs). Adopting the instrumentation reported in [14], we employ a foot step detection system using Interlink Electronics' Force Sensing Resistor (FSR) technology, shoe insoles, and an electrical circuit incorporating the sensors. Two FSRs were taped to each foot pad, one on the heel and the other on the ball of the foot. When pressure was applied to the FSR the resistance changed causing a change in voltage in the output of the electrical circuit, which is read by the Treadport server. Once either foot sensor (heel or ball) reached a specific voltage value, the server generated a step signal and applied the force profile to the user. The width of the force profile was also adjusted by the time length of the step so

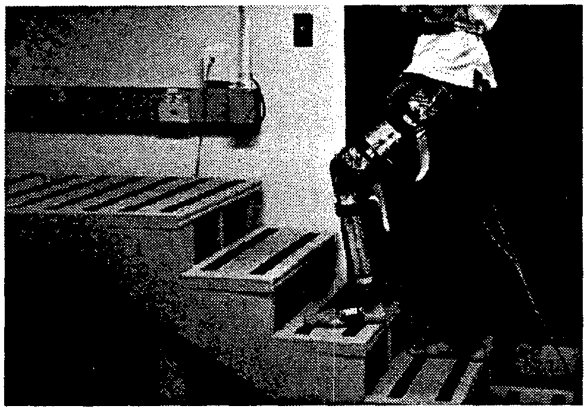

Fig. 3. Front side view of the stairs and sensor equipment used in stairs experiment

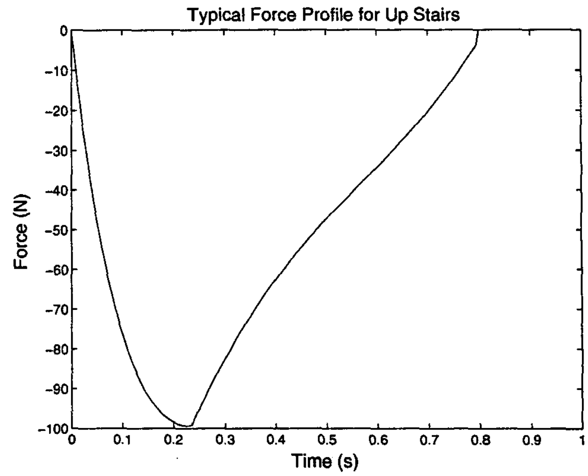

Fig. 4. Example virtual stair force profile.

that the force time would match the subject's step rate.

III. EXPERIMENTS

We conducted experiments with real stairs and the Treadport virtual stair simulation to determine how closely the subject's biomechanics during the simulation matched that of real stairs. There were 14 subjects, eight men and six women, most of whom had prior experience walking on the Treadport.

The real stairs consisted of five steps with a step height of 18.2 cm and a step width of 26.8 cm. A Northern Digital Optotrak 3020 3D motion tracking system was employed to calculate the segmental angles. Eight LED markers split into four groups of two markers each traced the foot, calf, thigh, and hip. The two markers are placed on a body segment using a wooden board with straps to wrap around the body and hold them in place. Figure 3 shows a subject standing on the stairs wearing the sensor equipment. The first part of the experiment is reading the sensor markers while the subject is standing straight; this defines the zero segment angles. For all Optotrak recordings the frame rate was 60 Hz, and data was recorded for 6 seconds. Subjects then walked up the stairs at least six times, three times starting off with each foot. Next the equipment was moved to the other leg (for marker visibility reasons), and the whole process was repeated for walking down stairs.

The data from the stair experiment was used to create an individual's force profile using the formula $f_{torque} = mgcos(\theta)$. The value $uh - sh$ is the distance from the subject's center of gravity to the top of the next stair. The subject's center of gravity is assumed to be at the center of the hip sensors. The distance w is estimated from the middle of the foot sensors. The force profile starts when the subject begins to lift off onto the next step and ends when the subject starts the next step. Note also that once the subject's center of gravity passes over the support point the force will become zero and then start to go negative as they prepare for the next step. We set the negative force portion to zero for the profile. The last step is to filter the beginning of the profile which otherwise would jump from zero force to a large force instantly. For this a simple exponential filter is used. Figure 4 gives an example of a finished force profile for going upstairs which has been normalized to a width of one second and an amplitude of 100 N. Force profiles were also calculated for going downstairs using the same principles.

To make the simulation more realistic, a visual display of the stairs to a Mayan temple was created. We also added a bouncing motion to the eye position, using data from the stair trials as a model, instead of basing the eye elevation purely on the stair model or an underlying slope.

The individual force profiles were then employed in the virtual stair experiments, which were broken up into three parts: general Treadport training, virtual stair training, and virtual stair experiments. The general Treadport training allows the subject to get comfortable manipulating the belt speed and handling the tether force; around 5 minutes sufficed. During the virtual stair training, the subject was given two force profiles - their individual force profile and a standard profile, chosen by a few expert users in pretrials. Each subject chose which profile they preferred and also adjusted the amplitude of the profile to an acceptable strength.

One other aspect to the experiments was to compare various slope forces to the virtual stair force to test that the virtual stairs were not in fact just simulating a specific slope. Thus the experimental trials included a full virtual stair force profile, a few slope forces, and two intermediate profiles composed of a percentage of the virtual stair force and the equivalent slope force. Instead of relying solely on a mathematical "equivalent" slope (constant) force, which is just the average of the force profile, we instead had each

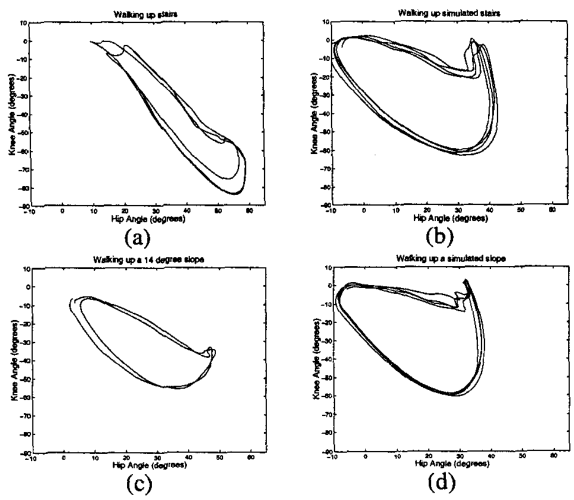

Fig. 5. Hip-knee cyclograms for four motions: (a) walking up real stairs; (b) walking up simulated stairs; (c) walking up a real slope; and (d) walking up a simulated slope.

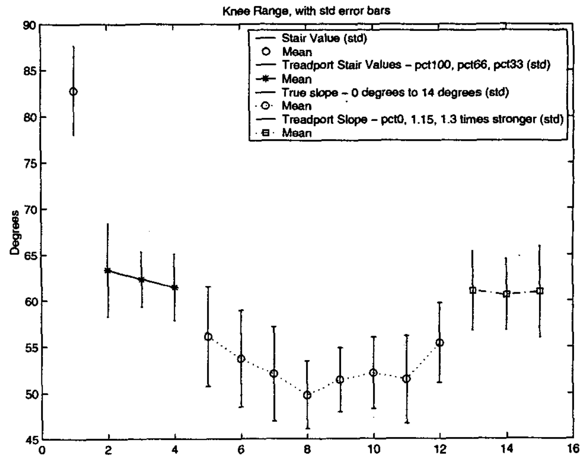

Fig. 6. Knee angle range over all experiments and subjects

subject determine the equivalent slope force in the training period. This was done by switching on the subject's command between the chosen stair force profile and a constant force, which they could increase or decrease, until the subject thought they both felt the same strength. These steps were also done for going downstairs.

Once completed, the actual Treadport trials started. Each subject performed 12 trials, 6 going upstairs and 6 going downstairs. The 6 force functions were full stair profile, 67% stair plus 33% equivalent slope, 33% stair plus 67% equivalent slope, full equivalent slope, 15% increase in equivalent slope, and lastly 30% increase in equivalent slope. The trials switched between going upstairs and downstairs with the order of the force functions randomized to reduce any learning affects and other biases. We started collecting the motion data after the user had taken a few steps with the specific force function and collected two sets of data for a total of 12 seconds. After each trial the subject was asked to give a psychological rating from 0 to 10 of how closely the experience felt to real stair walking.

After completing the experiments, we converted the 3D positions of the two sensors per body segment into lines with which we could determine the angles in between all four body segments. The angles were also adjusted according to the zero angle positions taken before the trials started. The next section gives the results of the experiments in terms of these body angles, absolute segment angles, and analysis of the hip-knee cyclograms.

IV. Results

The results consist of an analysis of four parts: (a) the real stair values obtained from the stair experiments, (b) the simulated stair and (c) the simulated slope values from the Treadport experiments, and (d) the real slope values obtained from previous experiments [12]. The analysis particularly focuses on how well the simulated stairs improve over the simulated slope when compared to the real stairs. We added the real slope values to check the validity of the simulated values. Figure 5 shows typical hip-knee cyclograms for all four types of motion in the up slope and upstairs direction. Much of the analysis is based on features of the cyclogram and its two parts, knee and hip angle. Also, all our analysis refers to upstairs and up slope as the biomechanics for going up differ more than for going down and our upstairs simulation performed better than our downstairs simulation.

As stated earlier the most striking difference in slope walking and stair walking is the larger knee angle range in stairs. Analysis over all subjects showed an average increase in knee range of 2.25 degrees in simulated stair walking compared to the equivalent simulated slope walking. The standard deviation was 3.03 giving a t-value of 2.774 over 14 subjects, which falls in the 99% confidence range. Thus, with 99% confidence the simulated stairs creates a larger knee angle, as in real stairs, over simulated slopes. Unfortunately though, while the increase is significant, it is rather small considering the average knee range for real stairs is over 20 degrees more than it is for Treadport slopes, creating only 10% of the ideal increase we would like to produce. Figure 6 shows the average knee angle range and standard deviation over all subjects and experiments. The figure is useful for showing the overall affect of the different forces and environments with respect to knee angle, but the analysis must be performed intra-subject.

Table 1 gives a summary of the means and Table 2 gives the t-value and confidence value for all of the variables. As the table shows, the simulated stairs are closer to the real

TABLE I

SUMMARY OF VARIABLE MEANS OVER ALL SUBJECTS AND EXPERIMENTS (UNITS IN DEGREES EXCEPT FOR ECCENTRICITY)

Variable	Stair Mean	Simulated Stair Mean	Simulated Equivalent Slope Mean	14 degree True Slope Mean
Knee Range	82.83	63.34	61.09	55.43
Hip Range	49.62	48.34	45.57	50.98
Calf Theta	68.28	70.93	72.40	73.88
Thigh Theta	156.20	172.87	173.76	161.17
Upper Body Range	6.47	9.41	7.38	3.54
Stance Angle	6.89	21.43	26.85	12.89
Knee Value at top of Cusp	-53.57	-18.81	-15.21	-37.08
Eccentricity	26.54	3.89	3.74	7.99
Separation at Stance	8.72	19.68	16.71	12.92

TABLE II

SUMMARY OF T-VALUES AND CONFIDENCE VALUE OF SIMULATED STAIR OVER SIMULATED EQUIVALENT SLOPE

Variable	T-value	Confidence Value
Knee Range	2.774	99%
Hip Range	3.991	99%
Calf Theta	2.434	98%
Thigh Theta	1.153	85%
Upper Body Range	3.285	99%
Stance Angle	5.228	99%
Knee Value at Cusp	3.411	99%
Sep. at Stance	-2.178	97%

stairs than the equivalent simulated slope on many biomechanical measures. Hip range as was found in an earlier study [12] was proportional to the slope angle. The hip range for stairs is nearly as much as for a 14 degree slope and the simulated stairs almost matches this value while the equivalent slope falls short. Calf theta and thigh theta are the mean absolute angles of the calf and thigh segment, respectively. For calf theta the simulated stairs again come closest to the real stair value. For thigh theta, however, the simulated stair is better than simulated slope, but the true slope value is much closer to that of real stairs. The upper body range refers to the sway of the upper body during a gait cycle which one would expect to be larger for stairs than for a constant slope. Our results echoed this expectation with the upper body range of real stairs being almost double that of a constant slope. For the Treadport the increase also held true with a significant increase in upper body range in simulated stairs over that of simulated slope. However, both values were considerable higher than their real counterparts.

The two stance variables refer to the upper left portion of the hip-knee cyclogram when the subject is almost back to a standing position during their motion. The stance angle refers to the angle between the incoming and outgoing line at this point in the cyclogram (Figure 5), while the separation at stance is the perpendicular distance between the two lines. The stair cyclogram has a narrow stance shape with a small stance angle and separation, while the cyclograms for slopes have a more rounded shape. For stance angle, simulated stairs matched closer with real stairs than the simulated equivalent slope, but the true slope cyclograms were even closer. However, for the separation measurement both simulated slope and true slope matched closer to real stairs than our simulated stairs.

The eccentricity of the cyclogram, which is very large for stairs, was essentially the same for both Treadport simulations and much lower, while the eccentricity of the true slope cyclogram matched the stairs much better, but still fell quite short. One last aspect of the cyclogram is the cusp on the right side, which occurs from the foot hitting the ground causing the knee to flex under the strain. For stair walking the cusp occurs at a large knee angle since the foot lands on a higher step. While walking on a flat surface, we would expect less knee flexion when the foot lands which corresponds to the lower values for the Treadport simulations. The stair simulation does however cause a larger knee flexion at foot fall than the slope simulation, while the true slope outperforms them both.

We also analyzed the data with respect to gender and experience level on the Treadport. With respect to gender men and women generally followed the same patterns in their biomechanical motion. The most notable exception was in upper body motion in which women tended to lean much farther forward and not sway as much as men. Experience, on the other hand, seemed to have a much greater affect on the results. A total of three subjects were expert users, nine had some experience, and two were walking on the Treadport for the first time. Unfortunately, with the small data size for the expert and beginner set no concrete conclusions can be made. However, the expert users gave better results than the other two sets with respect to most variables, including cyclogram eccentricity, stance angle and separation, knee angle at cusp, calf range, and upper body motion. The biggest difference between the groups was in the upper body motion as experienced users had a much lower range of motion due to their experience with handling the tether force.

The results for the psychological ratings showed that almost all subjects thought the simulated stairs felt more like stairs than the simulated slopes. After normalizing the results the average rating for the simulated stairs was a 6 while for the equivalent slope it was a 4.6. In addition, by ordering the ratings based upon the progression from full stairs to full slope the results show a steady decline in the rating, ending at 4.3 for the strongest slope presented.

V. Discussion

As the previous section showed our simulated stairs were a closer match to real stairs than the simulated slopes in many respects. We were able to change the subject's biomechanical motion from a normal slope walking motion towards the motion of walking on stairs. Also, psychologically the subjects felt the simulated stair motion resembled that of stair walking. However, while the biomechanical motion was shifted towards stairs, the change in motion for many cases was not very large and still is far from equivalent. The most unexpected result was how well many of the variables analyzed came closest to the true stair value in the true slope experiments. The thigh theta mean, stance angle and separation, knee value at cyclogram cusp, and the eccentricity of the cyclogram were all closest to those of stairs in the 14 degree slope experiment. We believe this is because of two things. First, on the true slope the platform is angled up and thus when a subject steps forward their foot lands higher than their back foot as in stairs. In addition, their stride is naturally reduced and restricted with the slanted surface, which also occurs significantly while walking on stairs. The increase in height and constricted body motion create a more eccentric hip-knee cyclogram and increases the knee flexion when the foot lands. A flat treadmill does not restrict the user's lower leg motion on its own and their change in motion only comes from the applied tether force and the subject's compensation for the expected force.

We think with a few modifications the results could be improved by taking advantage of the true slope results. The current Treadport's tilt mechanism has not been implemented yet, but with either a constant tilt or some algorithm for varying the tilt over a step, in combination with the tether force, we could make a better stair simulation. The tether force would have to be adjusted to compensate for the slope effect that the tilt would create, but we would gain the difference in foot height and restricted movement that the tilt adds. In addition, the Treadport tether is only capable of generating a one dimensional force, but the resultant force profile generated by the stair force model is two dimensional. The tether position is also fixed and thus for many of our shorter subjects the applied tether force was actually in a slight upward direction, while the force model generates a downward force component. Adding the extra dimension would more accurately reflect our model and we believe improve the simulation. Another limitation in the applied force is its application through the harness. For a constant force the harness straps become taut and the force the subject feels closely matches the applied force, but for a variable force there is backlash in the system and the applied force is somewhat filtered when it reaches the subject.

In conclusion we feel our stair simulation on the Treadport made progress in simulating real stairs on a flat treadmill. In the future we hope to make the additional improvements in the mechanical tether and add tilt to our simulation method to further advance a recreation of stairs on a treadmill style locomotion interface.

Acknowledgments

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