Design and Validation of a Low-Cost Bodyweight Support System for Overground Walking

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Walking with bodyweight support is a vital tool for both gait rehabilitation and biomechanics research. There are few commercially available bodyweight support systems for overground walking that are able to provide a near constant lifting force of more than 50% bodyweight. The devices that do exist are expensive and are not often used outside of rehabilitation clinics. Our aim was to design, build, and validate a bodyweight support device for overground walking that: (1) cost less than \$5000, (2) could support up to 75% of the users' bodyweight (BW), and (3) had small (±5% BW) fluctuations in force. We used pairs of constant force springs to provide the constant lifting force. To validate the force fluctuation, we recruited eight participants to walk at 0.4, 0.8, 1.2, and 1.6 m/s with 0%, 22%, 46%, and 69% of their bodyweight supported. We used a load cell to measure force through the system and motion capture data to create a vector of the supplied lifting force. The final prototype cost less than \$4000 and was able to support 80% of the users' bodyweight. Fluctuations in vertical force increased with speed and bodyweight support, reaching a maximum of 10% at 1.6 m/s and 69% BW support.
[DOI: 10.1115/1.4047996]

Introduction

Walking with bodyweight support can aid gait rehabilitation and reveal insight into gait biomechanics and control. Providing bodyweight support reduces the mechanical demand on muscles and can make it easier to coordinate limb motion. For patients with limited strength, walking with bodyweight support essentially increases their strength, making it possible to practice walking. Bodyweight support systems typically provide a lifting force via a harness around the waist, thighs, and often the chest. The vertical lifting force counteracts the downward force on the body caused by gravity. These harness-based bodyweight support systems do not alter the weight or inertial properties of the individual body segments. Bodyweight supported gait practice can improve walking ability in people with Parkinson's, hemiparesis induced by stroke, or incomplete spinal cord injury [1–4]. Studies on neurologically intact human subjects have revealed bodyweight support alters the walk-to-run transition speed, reduces stance time, reduces metabolic cost of walking, and reduces some leg muscle activity while having no impact on other leg muscles [5–11].

The trunk has sinusoidal vertical excursion during gait, making it a challenge to provide a constant support force. Static bodyweight support systems do not adapt to changes in trunk height: they provide a set force when the trunk is at a nominal height [12]. When the trunk is moved above this height, the upward force is reduced or removed entirely. Many of these static systems make it very difficult to move the trunk below the nominal height. Static bodyweight support systems are relatively simple, but they restrict trunk movement and thus do not permit normal gait kinematics. Many bodyweight support systems are dynamic, in that they allow for vertical movement of the trunk. The fluctuations in support force vary between system designs. Minimizing fluctuations in support force is challenging and is mostly achieved with active controlled systems [13]. The benefit of providing a truly constant force is that it more accurately simulates a "reduced-gravity" environment. This is more important in research where the intent is to understand the effect of gravity on biomechanics than it is for clinical rehabilitation. The complexity of systems capable of minimizing force fluctuations is perhaps another reason why traditionally most bodyweight support systems used for gait rehabilitation have high fluctuations in support force across the gait cycle [13,14].

Walking with bodyweight support overground has more therapeutic benefits than walking on a treadmill with bodyweight support, but both have positive effects on walking ability [12,15]. Overground walking allows the person to traverse over obstacles, around corners, up/down small stairs, and to choose their own walking speed. The majority of bodyweight support systems are limited to treadmill walking, as it easier to implement a stationary lifting force than a mobile lifting force [5,7,16–18]. Although the majority of biomechanical studies on the effects of walking with bodyweight support are done using treadmills [7–9,17,19,20], a few studies investigate the biomechanics of bodyweight supported overground walking. The overground bodyweight support studies have not investigated the effects of walking at more than 50% bodyweight support [8,16,21–24]. To the best of our knowledge, the effect of high levels of constant bodyweight support on the biomechanics of walking at different speeds overground is currently unknown. Some patients with limited strength cannot easily takes steps without high levels of bodyweight support (around 75%) [25].

Dynamic overground bodyweight support systems with small force fluctuations have been developed but are generally expensive [14,26]. Excluding bodyweight support systems that roll along the ground (which cannot be used to walk over obstacles and have a large inertia which makes turning difficult), there are three commercially available overground dynamic bodyweight support systems: the Gorbel SafeGait 360 Balance and Mobility Trainer, the Bioness Vector, and the Aretech ZeroG (Table 1). These systems are fixed to the ceiling with a track and provide near constant force to the user. Only one commercially available system has provided data on its force fluctuations throughout the gait cycle. The ZeroG system (Aretech) reports peak errors of 7 lbs when providing a lifting force of 120 lbs [27]. The commercial systems have a very high price point of over \$200,000. While this price may be rationalized and worthwhile in a rehabilitation clinic where one system will serve many patients, these systems are expensive for biomechanics research laboratories.

It was our goal to develop a low-cost dynamic bodyweight support system for overground walking. We chose design constraints of a cost less than \$5000 and force fluctuations less than ±5% of bodyweight at normal gravity. We first present the design of the reduced gravity simulator and then show validation of the design goals.

Methods

Design. To provide a constant support force with few fluctuations we chose to use constant force springs (John Evans' Sons, Lansdale, PA). Constant force springs are coiled springs that

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Manuscript received January 14, 2020; final manuscript received July 10, 2020; published online September 3, 2020. Assoc. Editor: Elizabeth Hsiao-Wecksler.

Table 1 Comparison of bodyweight support of commercially available overground bodyweight support systems

Manufacturer	Device	Maximum dynamic support lbf	Maximum dynamic support kN	Maximum static support lbf	Maximum static support kN
Gorbel	SafeGait 360	225	1	450	2
Bioness	Vector	200	0.89	400	1.78
Aretech	ZeroG	200	0.89	450	2

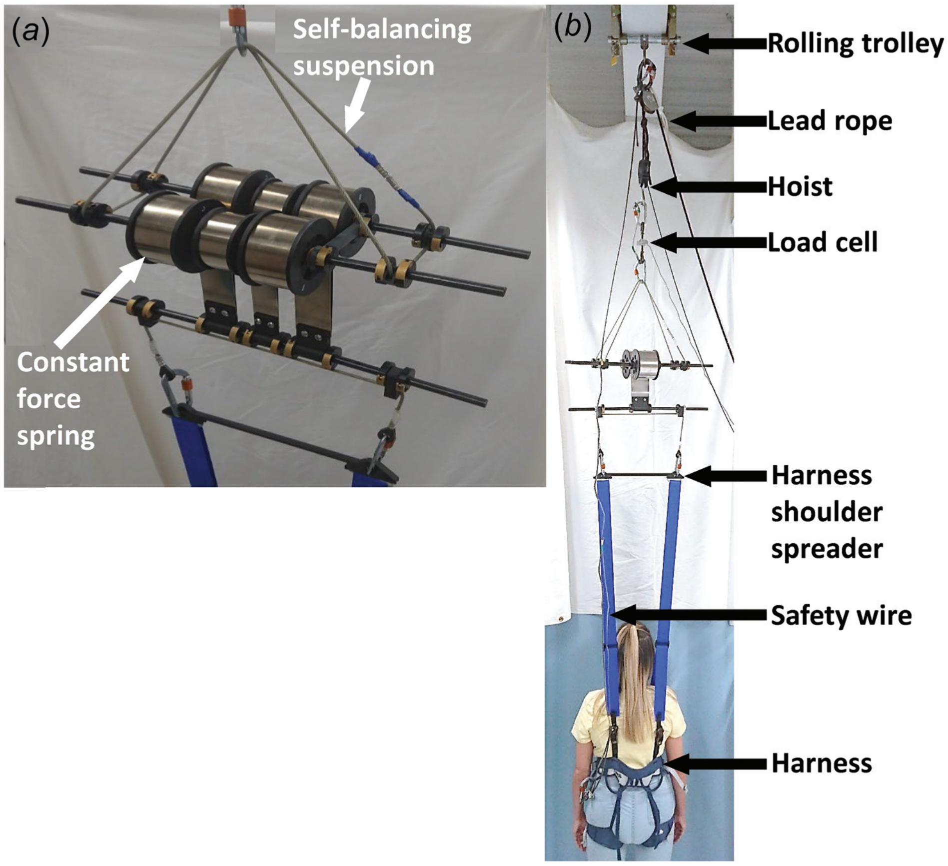

Fig. 1 The bodyweight support system we designed. Image (*a*) shows the main section of the system with three pairs of constant force springs. The entire support system, with just one spring pair, is shown in image (*b*).

when unraveled, provide a constant force regardless of the length of the spring uncoiled. We chose to use constant force springs because there are no traditional wire extension springs that could meet our design criteria of force fluctuations ±5% of bodyweight at normal gravity. Extension springs which have a low stiffness and could meet this criteria do not have a load capacity capable of withstanding the desired support force. Vertical center of mass excursion does not typically exceed 80 mm when walking [28–31]. For a 100 kg user, the design criteria define permissible force fluctuations as 98.1 N. Using Eq. (1), the spring stiffness must be less than ~1.23 N/mm (~7 lbs/in)

$$k = \frac{F}{\Delta x} \tag{1}$$

where F is maximum permissible force fluctuation, Δx is vertical excursion of the trunk, and k is the spring stiffness. Springs with this stiffness typically have a maximum load of less than 10 N (2 lbf), making them unsuitable for our design. We designed and built a system to suspend the constant force springs from a rolling trolley above a walkway, as shown in Fig. 1. We decided to suspend the system from a rolling trolley as it was an inexpensive and simple method of allowing the system to support the user in a three-dimensional (3D) space.

For stability, we mounted pairs of constant force springs back to back on 3D-printed spools with lips. The lips prevented the spring from moving on the spool and maintained the vertical direction of the unraveled section of the spool. The ends of the spring pairs were held together with carbon-fiber-reinforced 3D-printed parts (Markforged, Watertown, MA) and binding barrels (McMaster-Carr, Elmhurst, IL). The spring terminal and both spools were placed on $^1/_2$ in. (1.27 cm) diameter, 2 ft (61 cm) long, carbon steel rotary shafts (McMaster-Carr). Each spool could rotate freely about the rods. Spacer bars kept the rods apart, preventing friction between the spools in a pair. The spacer bars were a minimum of 5 mm longer than the center distance of the spools and were 3D printed from nGen material (ColorFabb, Belfeld, The Netherlands) at 70% infill. Easy-locking collars (McMaster-Carr) held all components on the rods. Needle rollers on either side of the spools reduced friction against other components (McMaster-Carr). We used coated wire rope (3/16 in. diameter with coating and 1/8 in. diameter without the coating, which is

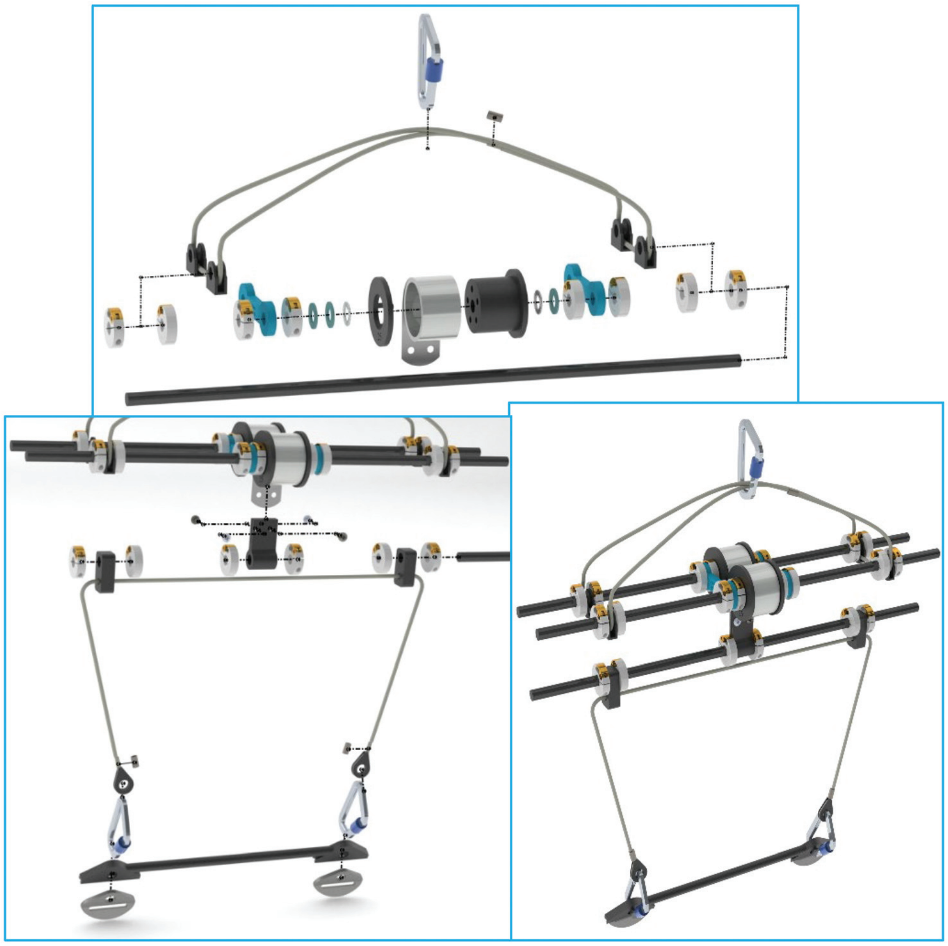

Fig. 2 Breakdown on the components used to build the main part of a single spring pair system. Top is the components that are slid and locked into place on each of the two top bars. Bottom left is how we assembled the bottom section of the system. Bottom right is the fully assembled spring system.

~47 and 32 mm, respectively) (McMaster-Carr) and 3D-printed guides (Markforged) on the far ends of each rod to suspend the system from a carabineer (MooseJaw, Madison Heights, MI). The rope suspension system was self-balancing and easily adjustable. We fixed a hoist (McMaster-Carr) to the rolling trolley (3 M DBI-SALA, Saint Paul, MN) on an I-beam above our overground walkway, which was connected to the carabineer attached to the bodyweight support system, with or without a load cell (Omega, Norwalk, CT) in series. A detailed breakdown of the spring section of the system is shown in Fig. 2.

To connect the bodyweight support system to the person, we modified a climbing harness (Petzl, Crolles, France) with two attachments for straps (Burton, Burlington, VA) on both the front and back. The front straps were attached using Nylon webbing (McMaster-Carr) and a carbon-fiber-reinforced 3D-printed bar (Markforged) to keep the two front straps level. Two pieces of Nylon webbing (McMaster-Carr) were looped through mounting plates (McMaster-Carr) and attached to the 4 attachment points on the harness. We designed and 3D-printed a carbon-fiber-reinforced bar (Markforged) that kept the left and right straps a set distance apart at the top, preventing the straps from touching the user's head. A safety wire of galvanized steel rope (1/8 in. thick, McMaster-Carr) was also connected between the trolley and the harness. A second, shorter safety wire was used in parallel with the load cell.

To provide a wide range of bodyweight support forces, our design was modular so that different constant force springs could provide different upward support forces. To be able to adjust support force with a good degree of accuracy, we used a variety of springs ranging from 4.4 to 40.9 lbs of force (~19.4 to 182 N). We chose springs with a cycle life of 2500 and above. We included high force springs in the design so that large support forces could be achieved with fewer components resulting in a low weight of the system, and straightforward assembly. To simplify balance, we kept spring components symmetrical about the middle, as shown in Fig. 1. For example, we could use a spring pair of force X in the middle, with matching pairs of force *Y* on either side. A nonsymmetrical system results in the system tilting and the resultant lifting force will not be completely vertical.

To engage the system, we used the hoist to stretch the springs. To allow for vertical movement in the system during gait, we stretched the springs 15 cm past their minimum engagement point. To maintain the position of the system above the user, we attached a rope to the rolling trolley that we used to pull the system along as the participant walked.

Validation. To validate our design, we asked healthy young human subjects to walk with bodyweight support while we recorded the support force. Before testing, the University of Florida's institutional review board approved the protocol and participants signed an informed consent form. We recruited eight participants, four of whom were female, with an average age of 27 years (±4, standard deviation), and an average weight of 682 N (±87). Participants walked at 0.4, 0.8, 1.2, and 1.6 m/s on an 8 m overground walkway and with 0%, 22% (±1), 46% (±2), and

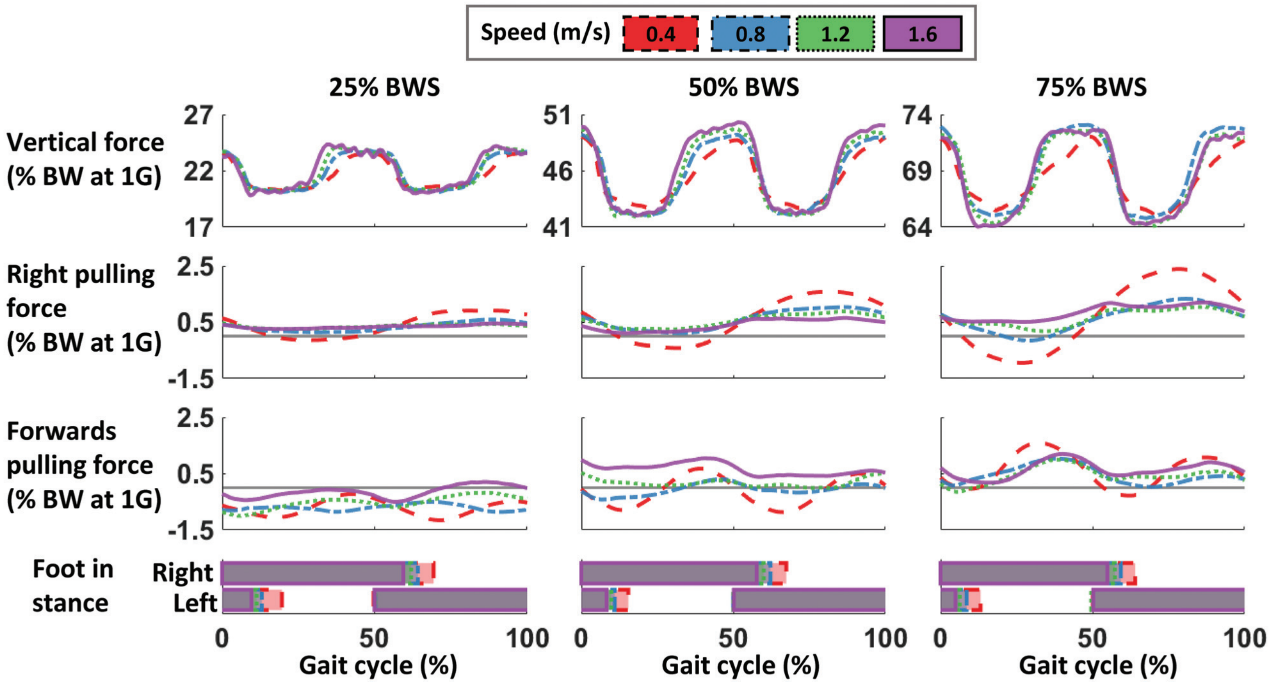

Fig. 3 Vector decomposition of the support force from the bodyweight support system over a stride at four speeds as found by loadcell measurements and kinematic positions

69% (±2) of their bodyweight supported. We chose to validate the design over a range of speeds and bodyweight levels because it is important to understand if the characteristics of the system are dependent or independent of speed and force. The fastest speed was chosen as it is a relatively fast walking speed for able-bodied people, and it was anticipated that most of our participants would be able to walk at this speed without having to transition to a running gait [6]. The maximum level of bodyweight support was based on previous studies walking with bodyweight support [13,25]. Further, we chose not to provide bodyweight support greater than 75% to minimize discomfort from the harness.

Force plates (AMTI, Watertown, MA) embedded in the walkway provided ground reaction forces and motion capture cameras and markers (Optitrack, Corvallis, OR) quantified the kinematics. We also used motion capture markers on the bodyweight support system to monitor its position and orientation during gait. To measure the force through the system, we used a loadcell in series with the bodyweight support system. For each combination of walking speed and bodyweight support level, we recorded four "good" trials wherein the right foot landed on only the first forceplate and no feet shared a forceplate for the duration of the stride. For each of the three bodyweight support system configurations, we also used the load cell to record the weight of the system when it was hanging.

We processed all data using custom written programs in MATLAB (Natick, MA) and VISUAL 3D (Kingston, ON). We filtered all data with a low pass fourth-order Butterworth filter with cut-off frequency of 10 Hz. We used an 18 N threshold on vertical ground reaction force data to identify heel strikes. We considered data from right heel strike to consecutive right heel strike as one stride and normalized all data to 0–100% of stride. If no ground reaction force data were available for the consecutive right heel strike, we used motion capture data processed via VISUAL 3D to identify heel strikes. To better understand the force provided by the system, we modeled the vector between the center of the four hip markers (right and left iliac crest and right and left posterior superior iliac spine) and the center of the bodyweight support system. We combined load cell measurements with this vector to determine the vertical force, forward/backwards pulling force, and sideways pulling force. Next, we subtracted the weight of the bodyweight support system from the vertical force to find the vertical bodyweight support force. To allow comparison across participants, we then normalized the support force to the participant's bodyweight at normal gravity (1 G). For additional insight, we calculated an effective bodyweight by summing the vertical ground reaction force of both feet during the stride and found the mean. We then found the normalized effective bodyweight support by dividing the effective bodyweight by the known bodyweight at normal gravity and subtracting the result from 1.

We averaged the data for each condition and each participant, before averaging across participants. To evaluate force fluctuation over a stride, we found the range of the bodyweight support force. We compared this range to our target fluctuations of ±5% bodyweight at 1 G. To confirm that our measurement of bodyweight support force from the load cell was accurate, we compared the normalized effective bodyweight support to the average of the normalized bodyweight support force.

Results

The cost of our modular system prototype was less than $4000. This cost does not cover installation of I-beam or purchase of 3D printers (Markforged X7 and Lulzbot Taz 6, Loveland, CO), but does include four pairs of each spring strength and enough rollers and collars for a four spring configuration. The quick release clamping collars were one of the more expensive components of the system. They could be replaced by a more affordable clamping collar (a two-piece set-screw collar, for example,) at the cost of increased time needed to switch or adjust components. The parts we designed and 3D printed have been uploaded to GrabCad (GrabCad Inc., Cambridge, MA) and are available to download [32].

The pattern of fluctuation in vertical and sideways force is consistent across bodyweight supported and speeds, while the forward pulling force is dependent on the condition. The force over a stride is shown in Fig. 3. Vertical lifting force is highest at the beginning, middle, and end of stride. The sideways pulling force is in the shape of a single sine wave. At the beginning, middle, and end of stride, the sideways pulling force is at its midpoint. When the participant steps on the right foot, the pulling force become more leftward and when the participant steps with the left, the pulling force is more rightward. The forward pulling force

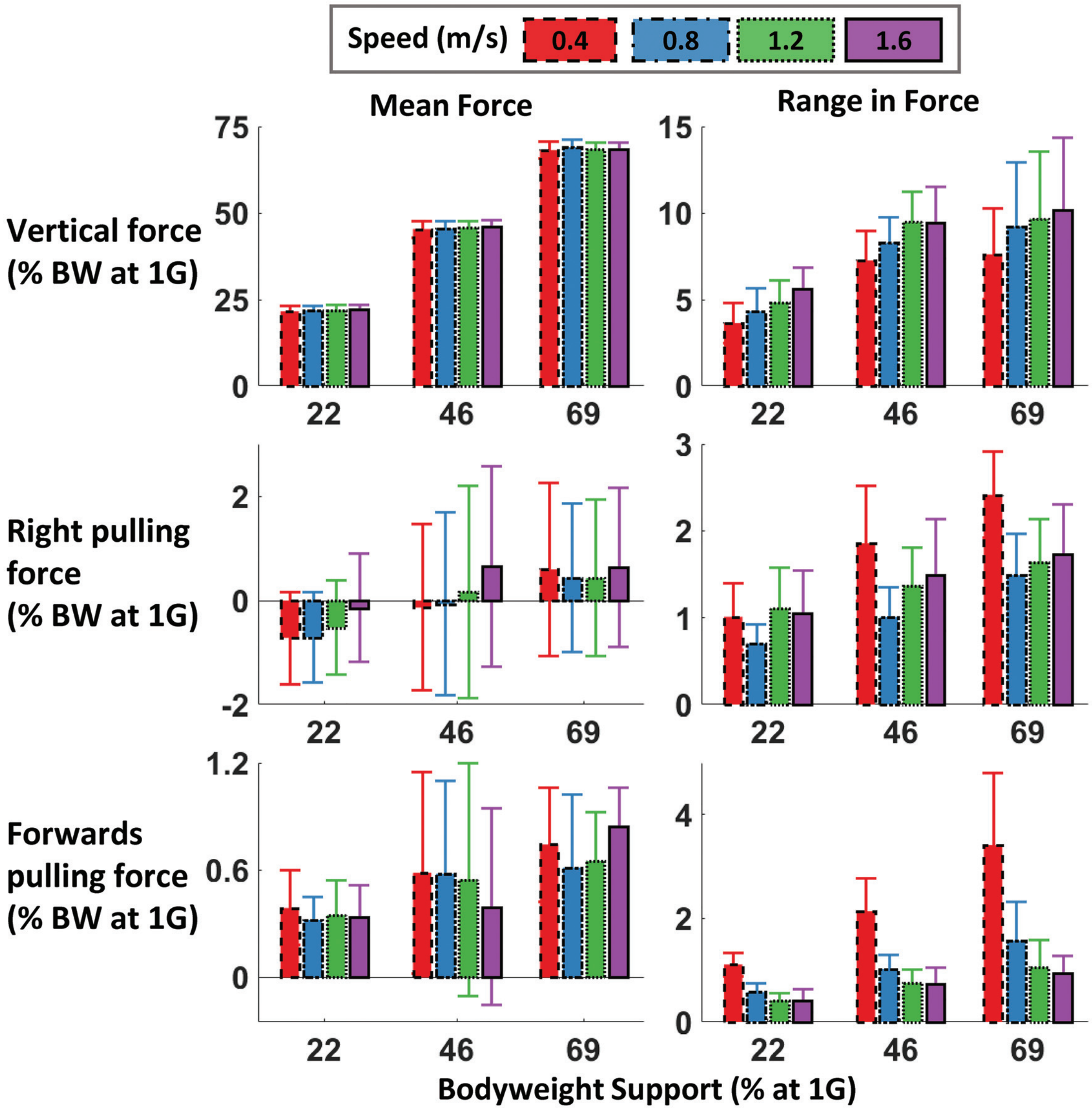

Fig. 4 Mean and range (with standard deviations) of support force over a stride at four speeds

is the most inconsistent across conditions, with greater variability at lower speeds and bodyweight support conditions.

The nonvertical forces provided by the system were very small (Fig. 3). The mean forward pulling force stayed within $\pm1\%$ BW at 1 G. Although the mean sideways pulling force somewhat favored pulling in the right direction, the mean force was less than 1.5% BW at 1 G.

The fluctuation in vertical, sideways, and forward force all increased with bodyweight support, as shown in Fig. 4. The vertical fluctuations also increase with increasing speed. The maximum vertical fluctuation was 10% BW at 1 G and seen in the 69% bodyweight support and 1.6 m/s condition. These data also confirm the system's ability to provide a vertical force of 79% BW at 1 G. The fluctuation in forward pulling force is highest at 0.4 m/s, but tends to increase between 0.8 and 1.6 m/s. The fluctuation in sideways pulling force tends to decrease with increasing speed.

The effective bodyweight support was similar to the vertical bodyweight support calculated from the load cell vector (Table 2).

Discussion

Our aim was to design a low-cost bodyweight support system capable of supporting up to 75% of the person's bodyweight, with vertical force fluctuations less than 10% BW at 1 G. Our prototype cost less than \$4000 and was capable of supporting 79% of the person's bodyweight. Force fluctuation in the forward and sideways directions were minimal (less than 3% BW at 1 G). The maximum vertical fluctuation did reach 10% BW at 1 G when the user walked at 1.6 m/s with high bodyweight support (0.31 G).

We believe that the vertical force fluctuations were caused by the vertical acceleration of the system and friction. The lower carbon steel rod and all the components on or beneath it, do move vertically to allow for the vertical movement of the users body. Because of the mass of the system, any acceleration of the moving section will translate into force. Further, it is possible that there was some friction between the interface of the spools and the rods on which they spun around. This frictional torque likely increases at faster speeds and when more force is transferred through the system. A later design will introduce bearings to the spools to determine if a reduction in friction reduces the vertical force fluctuations.

The sideways and forward pulling forces were close to zero, with very small fluctuations. The sinusoidal fluctuation in right pulling force is likely due to the person's center of mass moving from side-to-side to balance over the supporting foot. This sideways shift of the person is exaggerated at slower speeds when the participants have greater lateral center of mass excursion [33].

Table 2 Comparison of the difference in mean bodyweight support force found from load cell measures and vertical ground reaction force. The average bodyweight support forces were calculated by averaging the mean support force for each participant. The force found from the load cell and marker data was subtracted from the bodyweight support force found from the ground reaction force. Values are mean ± standard deviation.

		Difference in calculated bodyweight support force (% BW at 1G)		
		25 % BWS	45 % BWS	70 % BWS
Speed (m/s)	0.4	2.8 ± 4.2	−1.2 ± 5.4	−3.0 ± 5.1
	0.8	1.9 ± 4.8	−1.7 ± 5.6	−3.7 ± 5.3
	1.2	1.7 ± 4.5	−1.8 ± 5.3	−3.8 ± 5.4
	1.6	1.5 ± 4.1	−2.1 ± 5.5	−3.6 ± 5.5
	Mean	2.0 ± 4.2	−1.7 ± 5.2	−3.5 ± 5.1

The magnitude of these force fluctuations are very low and well within our acceptable range. Maintaining the position of the rolling trolley directly above the participant was relatively easy. By walking slightly ahead of the participant and ensuring that the lead rope was always taught, we were able to minimize forward pulling force to less than 2% of bodyweight at 1 G.

The bodyweight support measured by the load cell was very close to that found using vertical ground reaction force. Small differences between the two measurements can be attributed to three sources. The first is that the participant's bodyweight was measured before the additional mass of measurement instruments and harness were added. The second is that the ground reaction force calculation is sensitive to any change in velocity. If the participant sped up or slowed down slightly while on the force plates, this calculation is less accurate. The final reason for the small discrepancy could be due to the fluctuations in vertical support force over the gait cycle.

The maximum support force the system can provide is dictated by the hoist and the weight of the springs. We used a rope-pulley hoist from McMaster-Carr that was rated to 250 lbs (∼1.1 kN). The mass of the system below the pulley plus the support force must be less than 250 lbs. The mass of the system depends primarily on the number of springs used. The heaviest setup we used in this study was 16.3 lbs (72.6 N), meaning the system could not support more than 233 lbs (∼105 kg). A hoist with a higher load rating would increase the maximum support force. The rolling trolley limited the maximum user weight because it must be able to support the full weight of the user in the case of a fall. The trolley had a weight capacity of 141 kg (310 lbs). The maximum permissible support force the system could provide to a user of 140 kg is 75% bodyweight (equivalent to 105 kg of support). Additional consideration should be given to the possibility of bending the carbon steel rods. The lower rod experiences the most force, so we determined maximum support force using the maximum permissible bending moment of the lower carbon steel rod used in this design with the following equation:

$$M_{\max} = \frac{\sigma_{\max} I}{r} \tag{2}$$

where $M_{\max}$ = maximum bending moment, $\sigma_{\max}$ = yield strength, I = moment of inertia, and r = radius. The maximum bending moment of the 0.5 in. (12.7 mm) diameter carbon steel rod was found to be 104 N·m (920 lbf in.). Equation (3) can then determine the maximum support force

$$F_{\max} = \frac{2M_{\max}}{L} = \frac{208\,(\mathrm{N}\cdot\mathrm{m})}{L\,(\mathrm{m})} \tag{3}$$

where $F_{\max}$ is the maximum user weight, and L is the position of one strap from the middle of the rod. If multiple springs are used, L will be larger which lowers the maximum support force. We did not induce any bending in the rod during our validation.

Individuals with severe gait deficits may experience slightly different fluctuation in support force compared to healthy controls. The fluctuation in support force may decrease or increase depending on the characteristics of the users gait. Gait with increased medial–lateral sway would likely see an increase in the medial–lateral support forces—but because of the long length between springs and person, this increase is small and should not be obviously noticeable by the user. A more lurching gait may make it difficult to maintain the position of the system over the users head. However, the walking speed of such gaits is expected to be slow, so it is realistic to assume that with practice, the person moving the system will be able to maintain it at an acceptable position and prevent large anterior posterior forces. The magnitude and speed of trunk vertical excursion will impact the vertical lifting force through altering the frictional force. If a person with gait difficulties has slower and smaller vertical movement of the trunk, the fluctuations in vertical support force are likely to be less than the fluctuations for healthy controls. The bodyweight support system should be able to compensate for a bent trunk. The straps that attach to the harness can be adjusted so one side is longer than the other, and/or the front straps are shorter/longer than the back straps. This ensures that the support force can be split evenly between all four straps and will not force the torso into an uncomfortable position. Validation of the force fluctuations in the system with abnormal gait should be performed.

There are two drawbacks of the system in comparison to the more expensive commercially available overground bodyweight support systems. The first is that the constant force springs used in the system do have a limited life span. The life span is the number of cycles a spring can complete before its force diminishes and the integrity of the spring decreases. This characteristic is determined by the spring's composition and can range from 2500 to 20,000. To ensure participants safety, we took note of the number of cycles completed and disposed of springs when they were close to reaching their life cycle. The springs undergo one cycle per step of typical gait, meaning the springs with the shortest life cycle can be used for 2500 steps. The springs range in cost from ∼$15 to $33, which makes the cost to maintain the system relatively low and affordable. The 3D-printed spools for springs can be reused, although the lids may need to be replaced if they cannot be removed as a single piece. The second issue is that our modified climbing harness was uncomfortable for some participants. This is a common problem in reduced gravity simulators used for research and is actually independent of the source of the upward force.

The overground bodyweight support system described in this paper was low cost and capable of supporting a large percentage of the participant's bodyweight at fast walking speeds with force fluctuations less than 10% of user's bodyweight at normal gravity. The design is fairly easy to replicate, and its modular design means it is straightforward to change the support force. We tested the system only on a straight walkway, but it could be used over steps, inclines or in standing to sitting tasks. Additionally, the system could be used over a treadmill as an alternate method of body-weight support for treadmill walking.

Acknowledgment

Han Nguyen and Angel Bu both made valuable contributions to the development of the device and assisted with data collection and processing.

Funding Data

- National Institute of Neurological Disorders and Stroke of the National Institute of Health (Grant No. RO1NS104772; Funder ID: 10.13039/100000065).

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