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Augmenting strength-to-weight ratio by body weight unloading affects walking performance equally in obese and nonobese older adults

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Introduction

The preferred walking speed of older adults is an important metric of health because it is predictive of survival, mobility limitation and disability, dependency, as well as hospital and nursing home admission (Laukkanen et al. 1995; Guralnik et al. 2000; Guralnik et al. 1994; Cesari et al. 2005). While the age-related decline in strength and loss of muscle mass are key determinants of walking performance (Visser et al. 2000), obesity is an independent risk factor that increases the probability for loss of function (Stenholm et al. 2007). Older adults who possess both obesity and muscle weakness have slower preferred walking speeds, greater declines in walking speed over time, and greater risk of developing a mobility disability than those with neither low strength nor obesity (Stenholm et al. 2009; Bouchard and Janssen 2010). In the USA, 78 % of male and 69 % of female older adults are overweight or obese, which has increased by more than 20 % in 10 years (Flegal et al. 2010). The combination of a high prevalence of obesity, low participation in muscle strengthening activities, and an expanding older adult population, is likely to increase the number of older adults who are physically disabled.

To be independent, older individuals must be able to move about their homes, bathe, transfer, dress, feed, and use the toilet without the assistance of others (Katz et al. 1963). These daily activities are of short duration, have a low metabolic cost, and are dependent on the muscles' ability to generate the forces needed to move the body in opposition to gravity (Samuel et al. 2013). Recent work in our laboratory has shown that reduced physical performance in overweight and obese older adults is related to a low strength-to-weight ratio (S:W) that exists as a result of excess fat weight (LaRoche et al. 2011a). Overweight and obese older women in that study demonstrated 24 % lower S:W across lower-extremity muscles, 38 % lower rate of torque development, and nearly 20 % slower maximal walking speeds than normal weight women. In those with low S:W, slow walking speed has been shown to be related to lower contact forces between the foot and ground that limit the ability to change momentum and accelerate the body against gravity (LaRoche et al. 2011b). Thus, the capacity to increase walking speed is proportional to the ability to exert force relative to body mass, making S:W a key determinant of physical function.

A second problem with possessing low S:W is that activities of daily living, including walking, are performed at a high percentage of functional capacity (Samuel et al. 2013; Hortobagyi et al. 2003). Research by Hortobágyi and colleagues demonstrated that older

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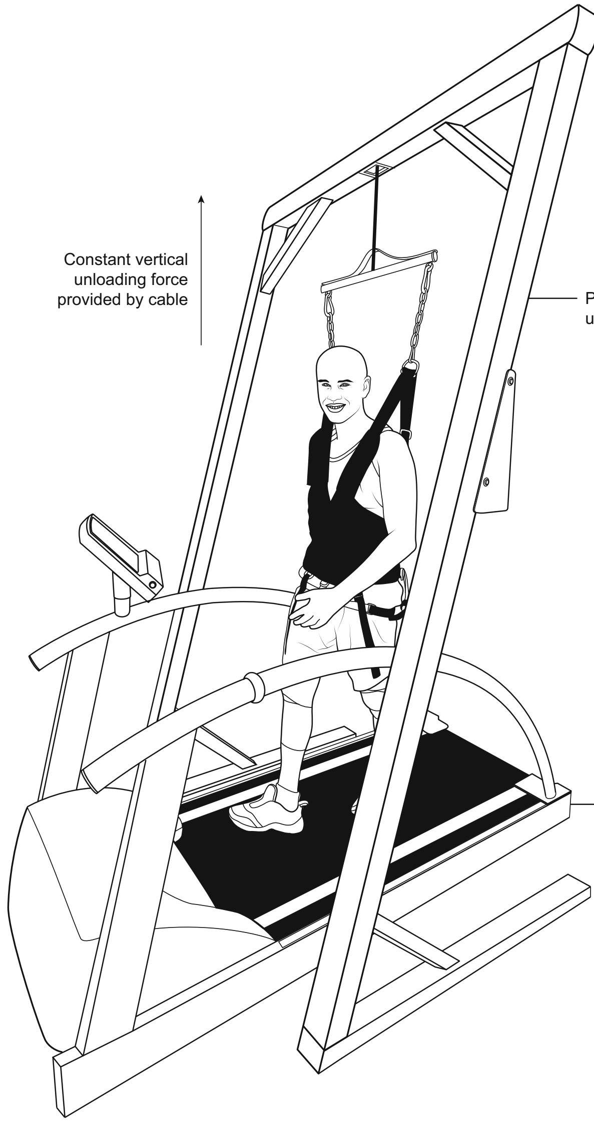
adults perform chair rises, stair ascent, and stair descent at approximately 80 % of maximal isometric leg press strength whereas young require approximately 50 % of maximal strength (Hortobagyi et al. 2003). Similarly, Samuel et al. demonstrated that knee extensor and hip extensor strength demand during walking in older adults were greater than 100 % of the available maximal isometric strength at these joints (Samuel et al. 2013). In older adults with low S:W, walking speed may be partially governed by the perception of effort in direct proportion to strength demand and muscle activation. Perceived muscular effort is thought to be mediated by the group III and IV afferent neurons that code information about muscle force, pain, and the metabolic state of the tissue in proportion to muscle workload (Laurin et al. 2015). Because of this, our central hypothesis is that if S:W is increased, older adults who are limited by low S:W will self-select faster walking speeds at similar levels of muscle activation and perceived effort.

During walking the muscles at the hip, knee, and ankle work synergistically to provide the supportive and propulsive forces necessary to elicit functional gait speeds (Winter 1980). Kinetic differences during walking exist at each lower-extremity joint between old and young, obese and normal weight, and older fallers and nonfallers that demonstrate reduced capacity to develop joint torques and powers relative to mass, and a reorganization of neuromuscular function between joints (Ko et al. 2010; DeVita and Hortobagyi 2003; Kerrigan et al. 2000). Despite the well-identified proximal shift of lower extremity power from the ankle to the hip in elderly gait (McGibbon and Krebs 1999; Graf et al. 2005; DeVita and Hortobagyi 2000), knee extensor S:W is important because it is the most routinely utilized metric of lower-extremity strength in the literature, particularly in large-scale epidemiological studies that characterize functional decline with aging (e.g., Health ABC, InCHIANTI, NuAge). Furthermore, knee extensor S:W criteria that quantify mobility limitation risk in older adults have been identified and validated (Manini et al. 2007b), whereas S:W criteria for hip extensor and ankle plantarflexor joint actions have not been well-established. Therefore, it is logical to use a knee extensor model to evaluate the relationship between S:W and walking performance, not because of a predominant reliance on the knee extensors during elderly gait, but so it may contribute to the existing strength and mobility literature and help explain some of the discrepant findings therein.

Knee extensor S:W cutpoints that identify risk for mobility limitation vary slightly between studies but generally indicate that a unilateral knee extensor S:W below 1.3 Nm kg^{-1} for women and below 1.7 Nm kg^{-1} for men, or an average of approximately 1.5 Nm kg^{-1}, increases risk for mobility limitation (Manini et al. 2007b; Ploutz-Snyder et al. 2002; Cress and Meyer 2003). In cross-sectional studies, there is a linear relationship between knee extensor strength and walking speed such that those with higher strength self-select faster preferred and maximal walking speeds (Rantanen et al. 1998; LaRoche et al. 2011b). Some long-term interventional studies have demonstrated that improving strength, decreasing body weight, or a combination of the two, result in improved mobility (Villareal et al. 2011; Brandon et al. 2003; Messier et al. 2004; Davidson et al. 2009; Anton et al. 2011) whereas other studies have demonstrated an inability of resistance training to improve mobility in older adults (Manini et al. 2007a; Mangani et al. 2006; Skelton et al. 1995). A research model that has the capacity to augment S:W acutely, in equal increments, may help identify the magnitude of strength gain needed to improve walking speed in older adults.

The primary purpose of the current investigation was to examine how manipulation of S:W affects self-selected walking speed and if acutely increasing S:W affects walking performance differently for normal weight, overweight, and obese older adults who possess different levels of S:W. To accomplish this aim, S:W was directly manipulated through a novel body weight unloading model (BWU, Fig. 1) that allows careful, incremental changes to S:W within subjects. Overweight and obese individuals are more likely to be below S:W mobility limitation risk thresholds, and thus it was hypothesized that they would experience the greatest change in walking speed as the muscular demands of weight support decreased. Similarly, it was expected that some of the age-associated and obesity-associated changes in spatial, temporal, and kinetic gait parameters (e.g., shorter strides, shorter single-limb support time, and lower supportive forces relative to weight) would improve to a greater extent in overweight and obese older adults when S:W was increased. This study may help elucidate the potential for changing walking performance in older adults through strength and weight loss interventions, and characterize muscle activation and gait changes during BWU treadmill walking that inform its therapeutic use in older adults.

Fig. 1 Experimental apparatus showing the relation of the walker, support harness, pneumatic body weight unloading system, and instrumented gait analysis treadmill



Materials and methods

Subjects

A random sample of older adults, age 65–85 years, was recruited from the community by newspaper advertisement. Twenty-seven men and women volunteered resulting in nine normal weight participants (body mass index, BMI = 18.5–24.9 kg m^{-2}, n = 5 women, n = 4 men), nine overweight participants (BMI = 25.0–29.9 kg m^{-2}, n = 9 women, n = 0 men), and nine obese participants (BMI > 30.0 kg m^{-2}, n = 5 women, n = 4 men). Volunteers were included if they were able to walk without the use of an assistive device, had no major disease or disability that prevented knee extensor strength measurement or treadmill walking, and received clearance from a primary medical care provider to participate in the study. The study was approved by the university's institutional review board for the use of human subjects, and all subjects gave their informed, written consent.

Procedures

Volunteers reported to the laboratory on three separate days, with 2 to 4 days separating visits. On visit 1, informed consent and measurements of height and weight were obtained, followed by completion of the Rapid Assessment of Physical Activity (RAPA) (Topolski et al.

2006) and the Short Physical Performance Battery (SPPB) (Guralnik et al. 1994). Participants were habituated to maximal voluntary isometric strength testing of the knee extensors on a commercial dynamometer (HUMAC Norm, *CSMI*, Stoughton, MA, USA), and S:W was determined from this initial assessment. The amount of body weight reduction needed to increase the knee extensor S:W by +0.1 Nm kg^{-1}, +0.2 Nm kg^{-1}, and +0.3 Nm kg^{-1} was calculated according to the algorithm described in Eqs. 1–3. Then, subjects completed a treadmill habituation session which included a 3-min warm-up walk at 0.8 m s^{-1}, walking for 1 min at maximal speed, and walking for 3 min each at four levels of S:W. At each level of S:W, volunteers were asked to self-select preferred walking speed as described in detail later.

During visit 2, knee extensor S:W was again measured and these S:W data were used for analysis in the study and final determination of BWU levels. Subjects completed another BWU treadmill walking habituation session at the four levels of S:W, again self-selecting preferred speed at each level. On visit 3, subjects were prepared for recording the electromyogram (EMG) from the vastus lateralis (knee extensor) and gastrocnemius lateralis (ankle plantarflexor) muscles of each leg and completed a self-selected maximal walking speed trial of 1 min duration to determine peak muscle activation. This was followed by two-minute bouts of walking under normal S:W, S:W + 0.1 Nm kg^{-1}, S:W + 0.2 Nm kg^{-1}, and S:W + 0.3 Nm kg^{-1} conditions in random order. Two minutes of seated rest were provided between each walking trial. At each level of S:W, participants self-selected preferred walking speed, and spatial, temporal, and kinetic gait parameters were obtained from the instrumented treadmill in the final minute of the walk, while EMG was measured via telemetry.

Strength measurement and S:W manipulation

The analog torque output of the dynamometer was imported into a data acquisition system (BIOPAC MP150, Biopac Systems, Inc., Goleta, CA, USA), was sampled at a rate of 1000 Hz, and smoothed by taking the mean of overlapping 50 sample epochs using the system's software (BIOPAC AcqKnowledge, Biopac Systems, Inc., Goleta, CA, USA). Subjects performed four maximal, isometric knee extensions, for each leg, for 5 s, with 30 s of recovery between efforts. The first two trials were used to refamiliarize the participant with the required maximal effort, and peak torque was recorded as the average of the last two trials. Peak torque was averaged between contractions and legs, and divided by body mass to obtain the unilateral knee extensor S:W (Nm kg^{-1}). A BWU algorithm was developed to determine the amount of body weight reduction needed to increase S:W in +0.1 Nm kg^{-1} increments as follows:

$$\text{Target S:W}\left(\text{Nm kg}^{-1}\right) = \text{Knee extenor S:W} + 0.1, 0.2, \text{or } 0.3\ \text{Nm kg}^{-1} \quad \text{(Eq.1)}$$

$$\text{Target Body weight (kg)} = \frac{\text{Knee extensor torque (Nm)}}{\text{Target S:W}\left(\text{Nm kg}^{-1}\right)} \quad \text{(Eq.2)}$$

$$\text{Target Unloading Weight (kg)} = \text{Subject's body weight} - \text{Target body weight} \quad \text{(Eq.3)}$$

The process was repeated for each subject to elicit the four S:W conditions. The weight reduction needed to manipulate S:W was provided by a pneumatic BWU system (PneuWeight, Pneumex, Sandpoint, ID, USA, Fig. 1). The BWU system applied an unloading force (controlled by microprocessor) to a full-torso harness worn by participants while allowing the normal vertical excursion of the body throughout the gait cycle. While the calculation of unloading weight was based on knee extensor S:W, it was expected that BWU also affected S:W for other lower-extremity muscles in a similar manner. Prior to walking performance testing, the target body weight was verified using the treadmill's integrated force plates. A sham unloading force of 9.8 N (1 kg) was used for the true S:W condition to partially blind the subjects to the specific S:W condition. Table 2 provides the target S:W, target body weight, and target unloading weight for each of the four S:W conditions by group.

Selection of maximal and preferred walking speed

Immediately prior to the BWU walking trials, with no weight support, maximal walking speed was determined by increasing treadmill speed by 0.015 m s^{-1} each second until the subjects reported that the speed was the fastest pace they could maintain for a short walk. Next, at each level of unloading, subjects began walking at a speed of 0.8 m s^{-1}, which based on our previous work was below the expected preferred walking speed. Preferred walking speed was then determined by increasing treadmill speed by 0.015 m s^{-1} each second until the subjects reported that the speed felt faster than they normally walked down the street to the store. Next, the speed was decreased in a similar manner until the subjects reported that the speed was slower than they walked down the street to go to the store. The process was repeated three times, and the three slow speeds and three fast speeds were averaged to obtain preferred walking speed. This method of preferred treadmill speed assessment has been shown to relate well to preferred overground walking speed. (Marques et al. 2013)

Gait measures

To measure gait performance at each S:W condition, participants walked on a motorized, instrumented treadmill (Gaitway II, Kistler Instrument Corp., Amherst, NY, USA) that was positioned directly beneath the unloading system (Fig. 1). The treadmill used in-deck force plates to record vertical ground reaction force (vGRF) and center of pressure data. These were then used by the treadmill software (Gaitway v. 2.0.8.50, Kistler Instrument Corp., Amherst, NY, USA) to determine kinetic, spatial, and temporal gait variables as described previously with very good reliability (LaRoche et al. 2011b; LaRoche et al. 2012). Kinetic variables included weight acceptance peak force (first peak of the vGRF) and push-off peak force (second peak of the vGRF); spatial variables included stride length and stride width; and temporal variables included stride frequency, single-limb support time, and double-limb support time. Each gait parameter was calculated independently for right and left feet over 20, sequential steps (ten strides), and the mean of these steps was used for analysis.

EMG

The surface electromyogram was used to assess the activation of the ankle plantarflexors and knee extensors. The skin was prepared to minimize impedance, and silver silver-chloride electrodes (Meditrace 530, Tyco Healthcare, Mansfield, MA, USA) were placed over the vastus lateralis and gastrocnemius lateralis of both legs according to Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM) standards. The EMG signal was obtained from the skin using preamplifiers (BN-EMG2, Biopac Systems, Inc., Goleta, CA, USA) at a gain of 2000× and was recorded by the data acquisition system at 1000 Hz. The signal was then bandpass filtered between 20 and 500 Hz, rectified, and integrated every 20 samples. The peak of the integrated EMG signal was recorded for each leg during the maximal speed walking trial and all subsequent EMG data were normalized as percent of peak EMG at maximal speed, and subsequently under each S:W condition were normalized to velocity to account for differences in walking speed between groups and conditions (Hof et al. 2002). The peak integrated EMG from the first five strides of the recording, for each leg, was obtained and averaged prior to statistical analysis.

Statistical analysis

Differences in subject characteristics and unloading parameters between BMI groups were compared using a one-way analysis of variance test. The Pearson product moment statistic was used to examine the relationship between strength and preferred overground walking speed. Repeated measures analysis of variance was used to compare the normal weight, overweight, and obese groups, across S:W conditions, for the dependent variables: self-selected preferred walking speed, weight acceptance peak force, push-off peak force, stride length, stride width, stride frequency, double-limb support time, single-limb support time, knee extensor activation, and ankle plantarflexion activation. The rejection criterion for the analysis of variance tests was $p < 0.05$. When significant main effects of group existed, Tukey's post-hoc test was used to identify the source of group differences. Independent and dependent t tests were used to investigate other significant main effects and interactions as appropriate. To control for multiple comparisons, the false discovery rate procedure was used for post-hoc t tests and the critical significance level was downwardly adjusted from $p < 0.05$ according to the methods of Curran-Everett (Curran-Everett 2000). All analyses were performed using a statistical software package (IBM SPSS Statistics, v. 20, IBM Corporation, Armonk, NY, USA).

Results

Subject characteristics

The three body weight groups were of similar age and height and reported similar levels of aerobic and strength activity (Table 1). Body mass and BMI were greater in the obese group, and obese subjects

Table 1 Participant descriptive characteristics

	Normal weight	Overweight	Obese
Age (year)	70.9 (6.2)	70.2 (4.5)	70.3 (6.1)
Mass (kg)	64.2 (11.7)	70.7 (7.3)	104.2 (24.0)[a,b]
Height (m)	1.66 (0.12)	1.63 (0.07)	1.70 (0.12)
Body mass index (kg m^{-2})	23.1 (1.5)	26.8 (1.5)	36.1 (8.5)[a,b]
RAPA aerobic activity score	5.7 (1.3)	5.3 (1.3)	4.9 (1.6)
RAPA strength activity score	1.7 (1.4)	1.2 (1.1)	1.3 (1.2)
Knee extensor strength (Nm kg^{-1})	1.82 (0.46)	1.43 (0.31)	1.41 (0.59)
SPPB total score	11.3 (1.4)	11.8 (0.4)	10.7 (1.0)
SPPB balance score	3.6 (0.9)	4.0 (0.0)	3.6 (0.7)
SPPB walking speed (m s^{-1})	1.44 (0.44)	1.27 (0.20)	1.13 (0.24)
SPPB chair rise time (s)	8.2 (2.1)	9.4 (2.1)	11.3 (2.5)[a]
Maximal walking speed (m s^{-1})	1.72 (0.41)	1.55 (0.26)	1.30 (0.37)[a]

Values are mean (SD)

RAPA rapid assessment of physical activity, *SPPB* short physical performance battery

[a] Difference between normal weight and obese, $p \leq 0.05$

[b] Different between overweight and obese, $p \leq 0.05$

demonstrated a 38 % longer time to complete the chair rise test than normal weight ($p = 0.017$). The knee extensor S:W in normal weight was 22 % greater than the overweight and obese groups but did not reach statistical significance ($p = 0.169$).

Walking speed

Knee extensor S:W was significantly correlated with the preferred overground walking speed obtained from the SPPB (Fig. 2A). The interaction between the BMI group and S:W condition for preferred walking speed was not significant ($p = 0.645$). However, a significant main effect of group ($p = 0.028$) showed that the obese group preferred speeds 20 % slower than normal weight ($p = 0.011$) and 16 % slower than overweight ($p = 0.043$) when averaged across S:W conditions (Fig. 2B). There was no effect of S:W condition on preferred walking speed for any group ($p > 0.5$).

Kinetic gait measures

When vGRF kinetic data were expressed in absolute terms (Newtons), the obese group demonstrated 42 % greater weight acceptance peak force ($p = 0.001$) and 46 % greater push-off peak force ($p < 0.001$) than normal weight (Fig. 3). As S:W increased, obese subjects had a greater reduction in push-off peak force ($p = 0.006$) than the other groups (Fig. 3). Across groups, there was a linear decline in weight acceptance peak force (15 %) and push-off peak force (23 %) from the normal S:W to the +0.3 Nm kg^{-1} condition (all $p < 0.001$). When kinetic data were expressed in relative terms (body weights), a significant interaction ($p = 0.015$) showed that an increase in weight acceptance peak force occurred in overweight (3 %) and obese groups (6 %), as S:W was increased, that did not occur in the normal weight group. Across groups, push-off peak force relative to weight decreased by 9 % as S:W increased ($p < 0.001$).

Spatiotemporal gait measures

For the spatial gait variables of stride length and stride width, there were no significant interactions, BMI group effects, or S:W condition effects ($p > 0.10$). When evaluating temporal gait measures, no significant BMI group by S:W condition interactions existed ($p > 0.50$). Obese subjects had 8 % slower stride frequency than normal weight ($p = 0.025$) and 12 % slower frequency than overweight ($p = 0.001$) (Fig. 4A). Single-limb support time was 7 % less in obese compared to normal weight ($p < 0.001$) and 6 % less than overweight ($p < 0.001$) (Fig. 4B). Double-limb support time was 27 % greater in obese compared to normal weight ($p < 0.001$) and 22 % greater than overweight

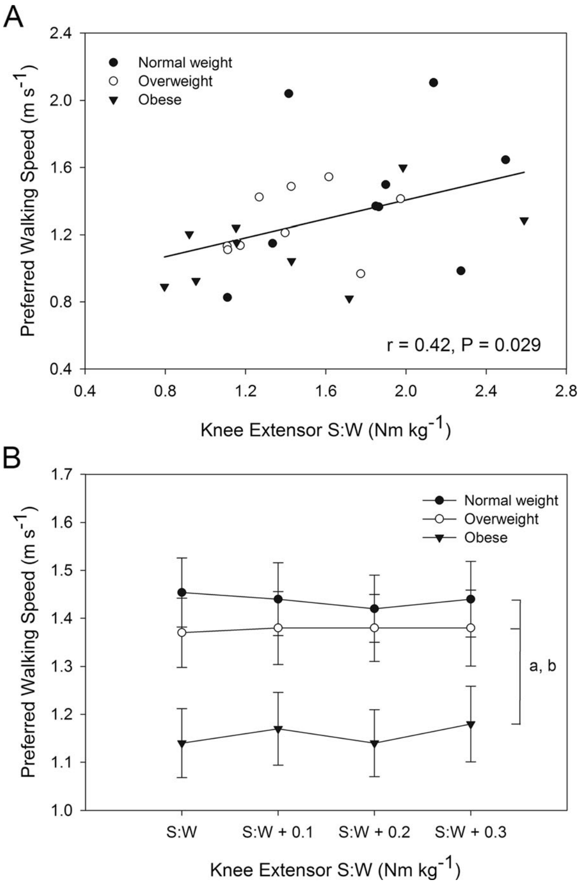


Fig. 2 Relationship between knee extensor strength-to-weight ratio (*S:W*) and preferred overground walking speed in a cross-sectional comparison among older adult participants (**a**) and comparing changes in preferred walking speed by weight group in response to acute increases in S:W of +0.1 Nm kg^{-1}, +0.2 Nm kg^{-1}, and +0.3 Nm kg^{-1}, which were elicited by body weight unloading (**b**). Panel B data are mean ± standard error. *a* significant main effect of group, different between normal weight and obese, $p < 0.05$; *b* significant main effect of group, different between overweight and obese, $p < 0.05$

($p = 0.001$) (Fig. 4C). When all participants progressed through the S:W conditions, single-limb support time increased 4 % ($p < 0.001$) and double-limb support time decreased 17 % ($p < 0.001$).

EMG

There were no significant interaction effects for muscle activation for either vastus lateralis ($p = 0.144$) or gastrocnemius lateralis ($p = 0.503$). A significant group effect existed for gastrocnemius lateralis activation ($p = 0.046$) due to obese subjects having 55 % greater activity than normal weight ($p = 0.014$). When all participants progressed through the S:W conditions, muscle activation of the vastus lateralis decreased by 9 % ($p = 0.012$) and gastrocnemius lateralis decreased by 14 % ($p < 0.001$) (Fig. 5).

Discussion

This study used a unique BWU model to test how augmenting S:W affects walking speed in older adults who differed by BMI. Overweight and obese groups demonstrated knee extensor S:W that placed them at risk for mobility limitation (Manini et al. 2007b), and the obese group had a preferred walking speed that was substantially slower than the normal and overweight groups. There was a weak positive correlation between knee extensor S:W and preferred overground walking speed, but contrary to our hypothesis, no change in preferred walking speed occurred when S:W was increased through BWU. As hypothesized, obese participants demonstrated slower walking speed, slower stride frequency, shorter single-limb support time, longer double-limb support time, and greater lower-limb muscle activation than normal weight and overweight participants. Increasing S:W by BWU improved gait timing and reduced muscle activation of the knee extensors and ankle plantarflexors, but this did not differ by BMI group.

Lower-extremity forces and preferred walking speed

The increase in knee extensor S:W used in this study was comparable to the knee extensor strength gains (0.3–0.4 Nm kg^{-1}) that occur as a result of resistance exercise programs in older individuals (Suetta et al. 2008; Cannon et al. 2007; Knight and Kamen 2001). However, when the range of S:W elicited by our BWU model is compared to the range of S:W possessed by participants in the study (nearly 2.0 Nm kg^{-1}, see Fig. 2A) the interventional increase in S:W may have been inadequate to affect preferred walking speed. Regardless, there is limited research identifying the magnitude of strength gain necessary to elicit improved walking performance in older adults. A recent review by Beijersbergen et al. (2013) demonstrated that in strength training studies, only 21 % of the variance in improved gait speed is explained by leg muscle strength, and concluded that the biomechanical mechanisms of how improved strength capacity transfers to walking

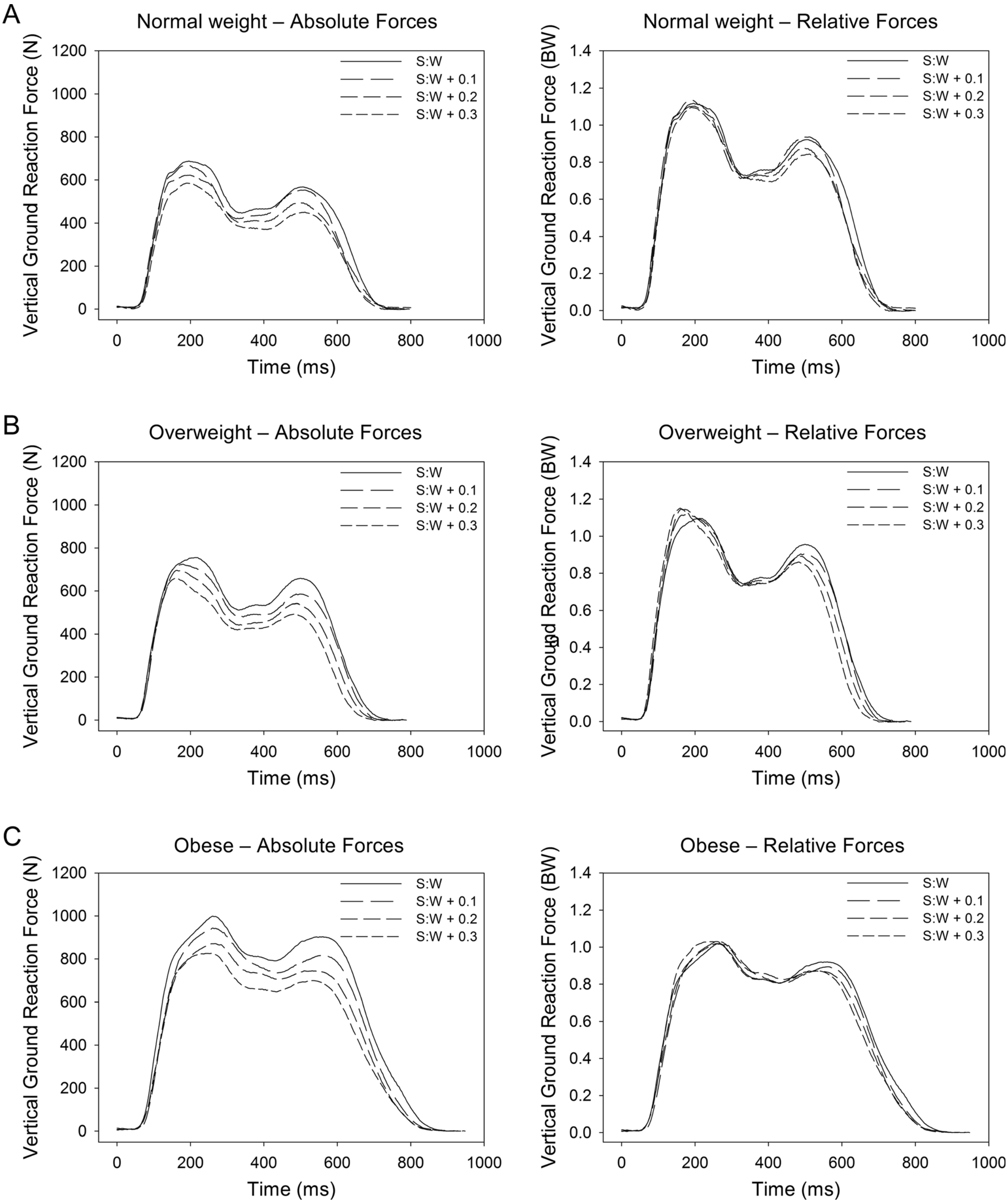


Fig. 3 Walking vertical ground reaction forces expressed in absolute terms (Newtons, *N*, *left panels*) and relative terms (body weights, *BW*, *right panels*) at the normal knee extensor strength-to-weight ratio (*S:W*) and at S:W +0.1 Nm kg^{-1}, +0.2 Nm kg^{-1}, and +0.3 Nm kg^{-1} for normal weight (**a**), overweight (**b**), and obese (**c**) older adults. Data are displayed as means

performance are unclear. A possible lesson learned from this study is that knee extensor S:W may need to be increased by more than 0.3 Nm kg^{-1} to affect the preferred walking speed of older adults.

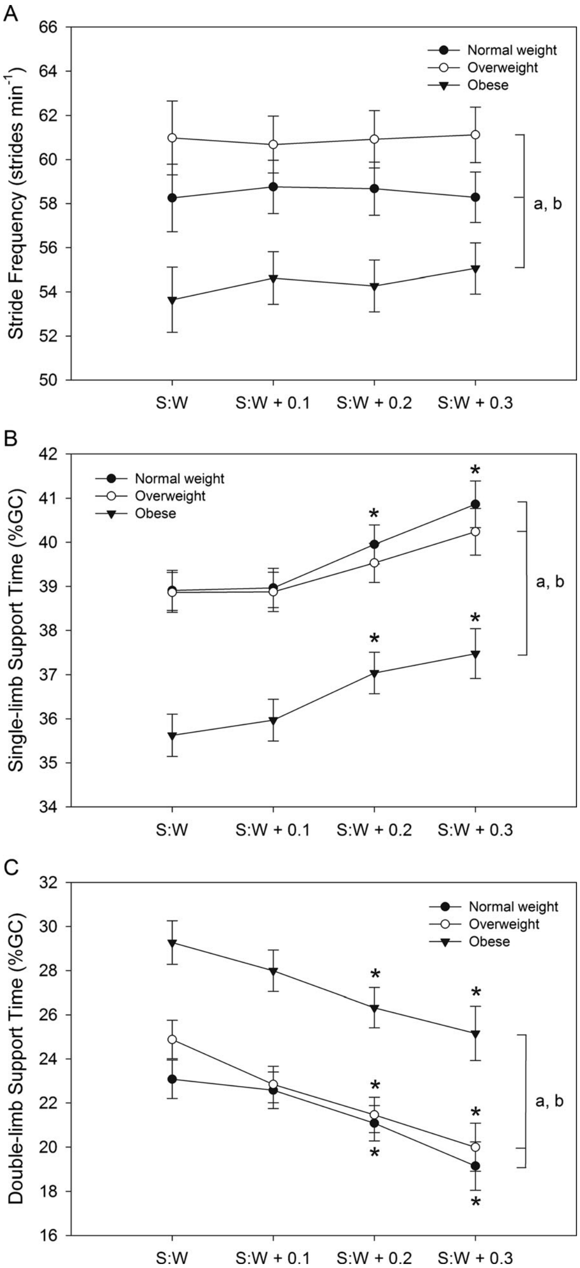


Fig. 4 Temporal gait data showing the effect of increasing knee extensor strength-to-weight ratio (*S:W*) by +0.1 Nm kg^{-1}, +0.2 Nm kg^{-1}, and +0.3 Nm kg^{-1} on stride frequency (**a**), single-limb support time (**b**), and double-limb support time (**c**) for normal weight, overweight, and obese older adults. *GC* gait cycle. Data are mean ± standard error. *a* significant main effect of group, different between normal weight and obese, $p < 0.05$; *b* significant main effect of group, different between overweight and obese, $p < 0.05$; *$p < 0.05$, significant change from S:W condition

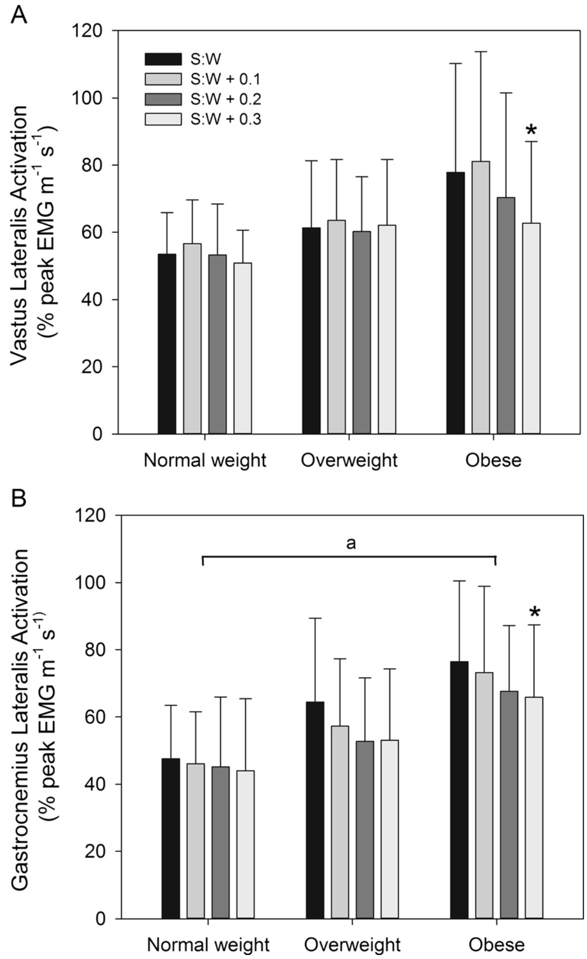


Fig. 5 Vastus lateralis (**a**) and gastrocnemius lateralis (**b**) muscle activation magnitude while walking with normal knee extensor strength-to-weight ratio (*S:W*) and at increases of +0.1 Nm kg^{-1}, +0.2 Nm kg^{-1}, and +0.3 Nm kg^{-1}, compared between normal weight, overweight, and obese older adults. Data are mean ± standard deviation. *a* significant main effect of group, different between normal weight and obese, $p < 0.05$. *$p < 0.05$, significant change from S:W condition

The vGRF is related to the sum of the hip, knee, and ankle extensor torques, represents the net propensity of the lower-limb to push away from the ground, and constitutes the largest component of the ground reaction force vector (Winter 1980). The vGRF acts in opposition to gravity to support body weight, its peak value relates to knee and ankle S:W in older adults, and the vGRF increases linearly with walking speed (LaRoche et al. 2011b). We theorized participants would maintain a constant muscular effort across S:W conditions, vGRF relative to body weight would increase as weight was reduced, and gait speed would increase concurrently. In contrast, the strategy that was employed by this study's participants in response to BWU was to reduce lower-limb forces while maintaining a constant speed, despite having the capacity to walk faster. As knee extensor S:W was increased via BWU, the absolute vGRF declined in proportion to weight reduction (Fig. 3 left panels) resulting in nearly identical vGRF relative to body weight across conditions (Fig. 3 right panels). Maintenance of an equivalent walking speed in the presence of reduced lower-extremity strength demand suggests that other factors, such as habit, balance confidence, effort, and cardiorespiratory capacity, may mediate preferred walking speed in older adults.

Muscle activation

Peak rectified EMG increases linearly with walking pace and thus speed must be considered when comparing EMGs obtained from different walking conditions (Hof et al. 2002). The muscle activation data in this study were normalized to speed to allow comparison between S:W conditions and BMI groups as walking speed differed between them. As such, the data portray what proportion of peak activation was utilized to walk at a speed of 1.0 m s^{-1}. With this interpretation, it is apparent that obese individuals require a higher level of motor unit activity of the lower-extremity muscles to walk. The consequence of compensating for low S:W through greater muscle activation is that muscular force cannot be sustained as high-threshold motor units fatigue more readily (Maffiuletti et al. 2007). Thus, by maintaining walking speed across S:W conditions, our older participants reduced muscle activation to a level that may help limit fatigue during walking, and appear to have prioritized reduced muscular effort over increased speed. This phenomenon could partially explain the lack of increased walking speed, despite strength gain, seen in some prospective resistance exercise studies.

Similar to the present study, Fischer and colleagues (Fischer et al. 2015) demonstrated approximately 30 % lower activation of tibialis anterior, lateral gastrocnemius, and vastus lateralis muscles in healthy, young participants when walking with 15 % BWU when speed was held constant. During BWU treadmill walking in older women, Thomas et al. (Thomas et al. 2011) showed no difference in vastus lateralis or biceps femoris activation between walking at comfortable speed (mean = 0.92 m s^{-1}) with 0 % body weight support and at fast walking speed (mean = 1.13 m s^{-1}) with 40 % body weight support. Thomas' findings highlight the importance of normalizing EMG data to speed as these results would likely have demonstrated a lower level of activation at the faster speed, 40 % BWU condition.

Spatial and temporal gait parameters

Walking speed is the product of stride frequency and stride length, both of which have been shown to decrease with age and obesity (Lai et al. 2008; Samson et al. 2001; LaRoche et al. 2011a). Increasing S:W via BWU did not affect stride frequency or stride length in any group, explaining the lack of change in walking speed. These findings are in agreement with Thomas et al. who showed no change in stride length or stride frequency with either 20 or 40 % of body weight unloading (Thomas et al. 2007a), yet van Hedel et al. showed a decrease in stride frequency and increase in stride length, but only at the most extreme level of BWU (75 % of body weight) (van Hedel et al. 2006). The maximal BWU in this study was only 18 % of body weight, and the lack of effect on these spatiotemporal variables may have occurred since the BWU model directly affected S:W, but did not affect inertial properties of the limbs. It is interesting to consider that the mass of the limbs, their moments of inertia, and the muscular work needed to accelerate them may explain the slower stride frequency seen in the obese subjects (Browning et al. 2007), and perhaps is another reason why when body weight was unloaded participants selected equivalent preferred walking speeds.

Temporal gait parameters are influenced by age, body anthropometrics, and walking speed. Double-limb support time occupies a greater proportion of the gait cycle when comparing old to young walkers (Winter et al. 1990), elderly fallers to non-fallers (Mbourou et al. 2003; Maki 1997), and when comparing obese to normal weight walkers (LaRoche et al. 2011a; Lai et al. 2008; Browning and Kram 2007). Spending more time in double-limb support is a strategy that may increase stability and allow a greater extent of the work of weight support to be shared between limbs. As some of the burden of weight support was removed to increase S:W, double-limb support time decreased and single-limb support time increased, mitigating a portion of the age-associated and obesity-associated effects on these variables. These gait cycle shifts occurred independent of walking speed as it remained consistent across levels of S:W. This suggests that qualitative improvements to older adults' gait, such as improved gait timing and balance control, could occur with increased S:W in the absence of increased speed. The changes in gait timing under BWU conditions are similar to those reported by van Hedel et al. (van Hedel et al. 2006) and Fischer et al. (Fischer and Wolf 2015) and suggest that deficits in the capacity for single-limb support may be partially mediated by S:W. Single-limb and double-limb support times are sensitive to changes in S:W, making these temporal gait variables potentially useful markers of gait quality in older adults.

Practical applications

The lower muscle activation and metabolic cost of BWU treadmill walking (Thomas et al. 2007a) may make it an appropriate adjunct exercise therapy for frail or obese older adults with low S:W, poor balance, mobility limitation, or orthopedic limitations who have difficulty walking for more than a few minutes without assistance. While a multitude of studies have used BWU treadmill walking after stroke, spinal cord injury, or in patients with cerebral palsy, multiple sclerosis, or Parkinson's disease, only a few recent studies have used BWU to combat the age-associated decline in mobility. Thomas et al. successfully used a BWU treadmill walking intervention to improve the maximal overground walking speed of older women by 13 % in comparison to a 3 % increase in controls (Thomas et al. 2007b), and Peterson and colleagues similarly demonstrated an 18 % increase in usual gait speed and 14 % increase in rapid gait speed compared to baseline measurements (Peterson et al. 2014). While reduced muscle activation with BWU may limit exertion and prolong endurance, lower activation could lead to neuromuscular detraining if it is used frequently in place of unsupported walking.

Although the prevalence of osteoarthritis was not evaluated in the current study, obese older adults are seven times more likely to have osteoarthritis in the lower extremity (Ackerman and Osborne 2012) and could benefit from exercising with lower joint forces. In this study, the absolute weight acceptance peak force and push-off peak forces were reduced by 13 and 23 %, respectively, in the obese group at the highest level of BWU (18.3 kg reduction, 18 % body weight). The reduced ground reaction forces that occur with BWU should elicit lower skeletal loads, possibly contributing to improved exercise tolerance in older adults with osteoarthritis or other orthopedic limitations. In fact, Messier and colleagues report that for each 9.8 N weight

Table 2 Target strength-to-weight ratios, body weights, and unloading weights

Condition:	S:W	S:W + 0.1	S:W + 0.2	S:W + 0.3
Target strength-to-weight ratio (Nm kg^{-1})				
Normal weight	1.82 (0.46)	1.93 (0.45)	2.01 (0.45)	2.13 (0.44)
Overweight	1.43 (0.31)	1.53 (0.31)	1.62 (0.31)	1.72 (0.31)
Obese	1.48 (0.58)	1.58 (0.58)	1.69 (0.59)	1.79 (0.61)
Target body weight (kg)				
Normal weight	62.7 (11.6)	60.2 (11.3)	57.6 (11.1)	54.5 (10.8)
Overweight	70.3 (7.1)	66.0 (6.9)	62.0 (6.0)	58.5 (6.0)
Obese	100.1 (22.1)^{a,b}	93.3 (20.7)^{a,b}	87.2 (20.0)^{a,b}	81.8 (19.2)^{a,b}
Target unloading weight (kg)				
Normal weight	1	2.5 (1.3)	5.2 (2.2)	8.2 (3.1)
% Body weight		3.9 %	8.1 %	12.8 %
Overweight	1	4.3 (1.0)	8.3 (1.6)	11.9 (2.0)
% Body weight		6.1 %	11.7 %	16.8 %
Obese	1	6.8 (2.6)^{a,b}	12.9 (4.0)^{a,b}	18.3 (5.1)^{a,b}
% Body weight		6.5 %	12.4 %	17.6 %

Values are mean (SD)

[a] Difference between normal weight and obese, $p \leq 0.05$

[b] Different between overweight and obese, $p \leq 0.05$

loss (1 kg), compressive forces at the knee decrease by 40 N, a factor of 1:4 (Messier et al. 2005).

Study limitations

The prime limitation of the current study is that it altered S:W artificially through BWU, which does not reflect how S:W would change in response to diet and exercise interventions. Even so, the model allowed for the study of how equal, incremental, acute changes to S:W affected walking performance and muscle activation in older adult groups who differed by BMI. A second limitation is that the amount of weight unloading needed to increase the S:W by +0.1 Nm kg^{-1}, +0.2 Nm kg^{-1}, and +0.3 Nm kg^{-1} was not equal between groups (Table 2), and likely explained the group by S:W condition interactions for the weight acceptance and push-off forces. These results do illustrate that if S:W is a limiting factor for walking performance in older adults, that it may take a greater magnitude of weight loss or strength gain to improve function in overweight and obese individuals than in normal weight. A third limitation is that this study did not provide a comprehensive analysis of lower-extremity kinetics and kinematics. While there is a linear relationship between vGRF and walking speed, it is the anteroposterior GRF that is responsible for forward translation of the center of mass. The BWU model reduced the demands of the limbs for vertical weight support, but did not alter the relationship between body mass and the anteroposterior GRF, which may help explain why walking speed was unaffected by weight unloading. A final limitation is that even though random sampling was conducted, the overweight group did not include any male participants. This may have affected some of the comparisons; however, most of the significant group effects were between the normal weight and obese groups which had an equal distribution of sex.

Conclusions

The cross-sectional data from this study show that a low S:W was associated with slower preferred walking speed but acute, artificial increases to S:W by body weight unloading did not change preferred walking speed of any BMI group. With increased S:W, participants reduced lower-extremity forces and muscle activation in order to maintain a constant speed, despite having reserve strength capacity. Increasing S:W did mitigate some of the age-associated and obesity-associated changes in gait timing, specifically, single-limb support time was increased, and double-limb support time decreased. BWU treadmill walking reduced lower-extremity ground reaction forces and muscle activation making it a potentially useful therapeutic modality for those with low functional capacity in conjunction with other training modalities.

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Compliance with ethical standards

Conflicts of interest The authors declare that they have no competing interests.

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