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Training implications of maximal forces on a computer-controlled and motor-driven leg press by age group, sex, footplate direction, and speed

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Abstract

Strength training that overloads lengthening muscle fibers may result in greater strength gains with less effort and perceived exertion than conventional training modalities. This study evaluates a device capable of this overloading (a motor-driven and computer-controlled leg press) to develop recommendations for future training interventions. Unimpaired younger and older men and women (7/group, total $n=28$) performed three maximal-effort trials for both directions of footplate motion (IN and OUT) at three speed profiles (knee rotation speeds of 15, 25, and 35°/s) on a motor-driven and computer-controlled leg press. Normalized forces were tested for effects of age group, sex, direction of footplate motion, and knee rotation speed. Peak forces were 57% greater for younger and 20% greater for IN. Trends of greater IN relative to OUT forces (IN overloading) were present in women, but this was due to an inverse correlation between strength and IN overloading that was independent of age group and sex. Leg press strength training on a device that is capable of overloading lengthening muscle fibers is a promising new training method that appears to have the greatest potential benefits for the weakest participants. Training target profiles on the device tested and others similar to it should be set based on participant-specific maximums across the ROM in both IN and OUT directions at a speed in the middle of the range to be trained.

Keywords

older adults; closed-chain; leg press; concentric; eccentric; dynamometer

1. INTRODUCTION

Strength and power decline with age (Whipple et al., 1987; Lindle et al., 1997; Metter et al., 1997) and these deteriorations, particularly of the lower extremities as measured by a leg press (Foldvari et al., 2000; Pijnappels, van der Burg, et al., 2008), are associated with limitations in functional mobility (Foldvari et al., 2000) and a fourfold increase in the odds

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of falling in older adults (Rubenstein and Josephson, 2002). Women have generally been shown to be weaker and lighter than men (Lindle et al., 1997; Metter et al., 1997; Jozsi et al., 1999; Reid et al., 2008). Older women, however, were not found to differ from older men in mobility disability despite this being predicted by lower extremity muscle mass and strength (Reid et al., 2008).

Progressive resistance training is an effective intervention for reversing age-related strength declines and improving physical functioning in older men and women (Liu and Latham, 2009). The resistance for this training typically comes from gravitational (weights), pneumatic/hydraulic damping, or elastic forces. These forms of resistance generally apply forces throughout the range of motion (ROM) that do not match the capabilities of the individual or exercise performed (i.e. joint mechanics; muscular length-tension and velocity-tension relationships). This could under-train portions of the ROM and result in sub-optimal training improvements. Additionally, the maximal resistance is generally limited by the tensile force that can be produced by shortening muscle fibers, which is most commonly called a concentric or “positive” contraction, but is more appropriately termed a shortening action (Faulkner, 2003). We will henceforth refer to the maximal force that can be produced by a shortening muscle action as “shortening strength”. Muscle fibers can also produce force when they are lengthening, which is most commonly called an eccentric or “negative” contraction, but is more appropriately termed a lengthening action (Faulkner, 2003). We will henceforth refer to the maximal force that can be produced by a lengthening muscle action as “lengthening strength”.

Both shortening and lengthening strength are required for functional activities, but peak lengthening strength is 4–64% greater than peak shortening strength (Porter et al., 1995; Rutherford et al., 2001) with lower perceived exertion and metabolic cost than required for shortening strength (Abbott et al., 1952; Lastayo et al., 1999; LaStayo et al., 2003). Furthermore, lengthening strength tends to be better preserved with age than shortening strength, even in individuals with chronic conditions (Raj et al., 2010; Roig et al., 2010). Because of this potential for greater muscular loading (and concomitant potential for greater strength gains) with less relative effort, training utilizing overloading of lengthening muscle fibers that can also adapt to the specific force-position and force-speed relationships of the exercise and individual is particularly attractive for older adults.

Resisting a powered cycle ergometer (lengthening-specific exercise) has been shown to significantly increase isometric knee extensor strength by 26% while conventional cycling (shortening-specific exercise) at the same percentage of maximum heart rate did not (LaStayo et al., 2000). However, this ergometer exercise differs from what is typically considered “strength training” (i.e. fewer repetitions performed at a significant percentage of one’s maximum effort). Lengthening-only strength training of a single joint on a conventional motor-driven isokinetic dynamometer generally has greater effects on lengthening strength while shortening-only strength training generally has greater effects on shortening strength (Duncan et al., 1989; Higbie et al., 1996; Hortobágyi et al., 1996). Training that overloads both shortening and lengthening muscle fibers results in more symmetric strength improvements (Colliander and Tesch, 1990; Connelly and Vandervoort, 2000; Valour et al., 2004). These previous findings suggest that any training program should

overload both shortening and lengthening strength in order to optimally train both directions of muscle action to maximize overall strength gains and transfer to functional improvement. As modalities employed for conventional strength are not capable of increasing resistance for the lengthening phase of an exercise, overloading of the lengthening phase has been underutilized due to the logistical and technical difficulties of applying the required variable resistance (Hortobágyi et al., 2001). Easy-to-use commercial devices are now and/or will soon be available that are capable of modulating target resistance by direction of muscle action and across the ROM, but the optimal procedures or processes for setting these resistance targets are not known.

Greater shortening forces are generated for slower speeds and greater lengthening forces are generated for faster speeds in isolated muscle preparations (Hill, 1938), but these effects have been difficult to reproduce in voluntarily activated human muscle (Enoka, 1996). The majority of these studies assessing volitional strength have operated at the single-joint level, but most functional activities involve multiple joints acting in combination. While training of multi-joint motions is common (e.g. leg or chest press), we are unaware of any study utilizing a motor-driven dynamometer system to test or train maximal shortening and lengthening strength across the entire range-of-motion of a functional multi-joint task. Here we present maximal strength testing data from such a system - a computer-controlled and motorized leg press capable of driving an instrumented footplate at specified speeds that are independent of the forces applied to it. This allows it to collect maximal force profiles both for the footplate moving away from the user (OUT) and moving towards the user (IN) at a specified speed. This system was used to evaluate the effects of age group, sex, direction of footplate motion, and knee rotation speed on footplate forces during a multi-joint leg press task.

Note that OUT and IN footplate motions may not perfectly correspond to shortening and lengthening strength for all muscles used for this task. The OUT and IN motions would correspond to shortening and lengthening strength for all uniarticular muscles used for a leg press (e.g. vasti, gluteals, soleus) or all muscles for a uniarticular exercise such as a knee extension, but this may not be the case for the biarticular muscles utilized for this closed-chain leg press motion. It is possible that these biarticular muscles (e.g. gastrocnemius, hamstrings, quadriceps, rectus femoris) may be held isometric or even shorten during the IN phase. Recent data has demonstrated this in the biceps femoris, where muscle-tendon units were found to lengthen slightly (<2%) in 36% of hop landings and 48% of drop landings when they would be expected to be lengthening based on joint motions (Oh et al., unpublished results). Thus, for the remainder of this paper we will only refer to OUT and IN footplate motions rather than shortening and lengthening strength.

Considering that the majority of functional activities in everyday life incorporate multi-joint rather than single-joint motions (DeLisa et al., 2005), assessments of multi-joint strength may be more representative of functional abilities than single-joint strength assessments. Moreover, while absolute strength is important for many functional activities that involve manipulating one's environment (e.g. lifting or carrying objects), for the purposes of mobility and fall avoidance and recovery one's own body weight is the only load of concern. In these cases, analyzing forces relative to the participant's own body weight is more

appropriate than testing the raw force data. Therefore, the purpose of this study was to determine if established single-joint effects of age, sex, direction of muscle action, and speed on maximal strength also apply to a multi-joint leg press in order to inform the design and implementation of future training protocols to improve fall recovery ability and functional mobility using this device and others similar to it. We hypothesized that bilateral leg press forces relative to body weight as measured on a motor-driven, closed-chain, quasi-isokinetic dynamometer would 1) be greater for younger adults, men, and IN phase, 2) decrease for OUT and increase for IN at faster knee rotation speeds, and 3) be greater for IN relative to OUT (i.e. better preserved) in the older participants.

2. METHODS

2.1. Participants

A convenience sample of 14 younger (18–35 years) and 14 older (>65 years) unimpaired men and women (Table 1) were recruited from the community for this laboratory-based study. All participants were screened by a physical therapist or geriatrician and found to have no major health problems or physiological impairment of any kind, 20/20 corrected vision, and good balance as defined by one-legged stance durations of over 30 seconds on each foot.

2.2. Instrumentation

A quasi-isokinetic, closed-chain, motor-driven, and computer-controlled leg press dynamometer (BLAST System; Bio-Logic Engineering Inc., Dexter, MI) was used for strength testing. This system can be used unilaterally, but all data were collected for bilateral testing. Feet were placed on the footplate with the heels resting on a safety interlock switchbar (footplate motion would stop if heels lift off this switch) shoulder-width apart. There is a similar interlock in the seat to prevent the pelvis from rising off the seat during a trial. The feet were not affixed to the footplate, but they were repositioned before every trial if they had been moved between trials. The high-traction surface of the footplate makes it virtually impossible to change the position of the feet during a trial. The starting position of the footplate and its maximum displacement were set automatically by the BLAST software based on the individual participant's height such that the knees started at approximately 90° of flexion and could extend almost completely. Actual knee extension in this device was limited by participant anthropometry (device assumes leg length based on height) and applied forces (compliance of backrest could result in greater knee extension for stronger participants). Knee hyperextension was prevented by a safety bar, which also served to customize footplate displacement for the individual participant. If a switch on the top of this safety bar was contacted by the popliteal region of the knee during the OUT phase the plate would stop; if it was contacted in the attempt to initiate the IN phase the device would gradually and incrementally reduce the maximum footplate displacement until the IN phase could be initiated without contacting the safety bar. Footplate motion was driven by a 3.7hp servo-motor and speed profiles were generated by the BLAST software using the manufacturer's proprietary methods to approximate a constant knee rotation speed, regardless of the forces applied to the footplate. The specific speed profiles used for each participant were calculated based on their height using a vendor-supplied table, so the

footplate moved faster for taller participants to attain the same knee rotation speed. Data for force normal to the footplate were collected at 80Hz by four uniaxial force transducers supporting the footplate corners.

2.3. Procedures

Testing took place on two days. After completing the informed consent process as approved by the local institutional review board on the first day, participant characteristics were recorded and a battery of clinical tests was performed. These tests included Functional Reach (Duncan et al., 1990), steady-state preferred and fast gait speeds across a 16-foot section of floor, 8-foot Timed Up and Go (Rose et al., 2002), Four Square Step Test (Dite and Temple, 2002), and Maximum Step Length (Medell and Alexander, 2000).

On a second day of testing, two hours of data collection for a parallel study were completed (results reported elsewhere). This data collection session served to warm-up the participants, but did not fatigue them. The participants were then seated in the leg press, instructed in its operation, and given at least one sub-maximal practice trial at each of the three speeds tested to gain familiarity with its operation. The speed profiles used approximated knee rotation speeds of 15°/s, 25°/s, and 35°/s. The order of presentation of these speed profiles was randomized and provided 30 seconds of rest time between trials within each three-speed block. Three repetitions of this three-speed trial rotation were performed with one minute of rest time provided between each rotation, for a total of nine trials (OUT and IN phase for each trial). Additional rest time was provided if requested.

When the participants indicated they were ready to proceed they began the nine maximal effort trials. For these trials, the participants were instructed to push as hard as possible against the footplate until it stopped moving. Each trial started when the participant began pushing on the footplate, which caused it to move OUT at a specified speed to maximum footplate displacement. After a brief delay (1–5 seconds) the participant was instructed to push on the footplate again, which caused it to move IN to the starting position at the same specified speed. Verbal encouragement was provided during and between trials (e.g. “OK, you got X last time, let’s see if you can beat that this time...”). Additionally, visual feedback was also provided in real-time by the force trace during the trial and by the display of peak OUT and IN forces on the screen after each trial.

2.4. Data Processing

Hypotheses were tested using force data that were normalized to body weight, although raw peak force data are also reported for reference. Since these single peak values do not necessarily represent the entire shape of the data curve, the force data were also examined using the areas under the force-time and force-position curves. These two area measures, which are respectively defined as “impulse” and “work” in physics or engineering, are effectively the mean force across trial duration and the mean force across footplate position. While these two areas are similar, the footplate position on the BLAST leg press is not perfectly linear with respect to time (quasi-isokinetic), thus there are differences between them. More importantly, the force-time areas by definition decrease with increasing speed (faster trials are completed in less time), while force-position areas are not directly altered

by speed (same distance is traversed by footplate at all speeds). This was corrected for by dividing the area under the force time curve by the time required to complete the trial. The greatest peak value and greatest areas under the curves of the three trials collected were used for each combination of direction of footplate motion and knee rotation speed.

2.5. Statistics

Effects of age group and sex on participant characteristics and clinical tests were tested using analysis of variance. As no hypotheses were posed for these variables, these data and uncorrected *p*-values were reported for comparison and discussion.

Repeated-measures linear mixed models were used to simultaneously test the effects of participant age group, sex, direction of footplate motion, and knee rotation speed on the dependent force variables. Our first hypothesis was tested using the effects of age group and direction of footplate motion on peak force and areas under the force-time and force-position curves. To better visualize and test the speed*direction and age group*direction effects, three additional dependent variables for the percentage of increase in the IN direction forces relative to the OUT direction forces (henceforth referred to as “IN overloading”) were calculated from the force variables where $\text{IN overloading} = (\text{IN value} - \text{OUT value}) / \text{OUT value}$ and tested as above.

Uncorrected *p*-values were reported for all effects $p < 0.05$, but all thresholds of significance for hypothesis testing were corrected as per Bonferroni for the number of independent factors (dependent variables were closely correlated. A priori main effects were considered significant at $p < 0.0125$ ($0.05 / 4$ independent factors)). SAS was used for all analyses (SAS Institute Inc., Cary, NC).

3. RESULTS

3.1. Participant Characteristics and Clinical Test Results (Table 1)

No hypotheses were posed for age group and sex effects on participant characteristics and clinical test results; these data are provided for reference and uncorrected *p*-values reported. The men were both taller ($p < 0.0001$) and heavier ($p < 0.0001$), which resulted in no sex effect on body mass index (kg/m^2). Older participants had greater weight ($p = 0.03$) and body mass index ($p = 0.004$) than the younger participants. No adverse events occurred during testing and it was well tolerated by all participants.

Preferred and fast gait speeds did not differ across any participant group. Younger participants had greater Functional Reach ($p = 0.04$) and Maximum Step Length ($p = 0.0009$) than older participants. Younger participants also completed the Timed Up and Go ($p = 0.03$) and Four Square Step Test ($p = 0.01$) faster than the older participants. While men had greater Maximum Step Length ($p = 0.003$), this effect was not significant when Maximum Step Length was normalized to participant height to account for the greater height of the men. Every correlation between performance on a mobility-related functional clinical test with a significant age group effect and any force measure was improved by normalizing the force measure to body weight (Table 2).

3.2. Effects of age group and sex on leg press forces (Table 3)

When the non-normalized force data were examined, men had significantly greater peak forces than women ($p=0.0004$) and the younger participants did not have significantly greater peak forces than the older participants ($p=0.004$). However, when the data were normalized to body weight the sex effect was no longer significant ($p=0.05$) and the age group effect became strongly significant ($p<0.0001$).

Younger participants also had greater areas under the force-time curve ($p=0.0003$) and the force-position curve ($p=0.0016$). IN forces were greater than OUT for all participants (peak and both area measures $p<0.0001$) and reached their peak values when the knees were more flexed as shown by representative data in Figure 1. Men had 43% greater area under the force-position curve than women ($p=0.0016$, 22.6 ± 8.1 vs. 15.8 ± 4.9 %body weight*m), but men did not have significantly greater areas under the force-time curves than women ($p>0.05$).

3.3. Effects of footplate speed on leg press forces (Table 3)

Knee rotation speed had no significant effect on peak force or the areas under the force-time (when corrected for trial duration) or force-position curves (Figure 2 and Table 3), although a trend of decreasing peak forces for faster speeds was present ($p=0.03$). While the area under the force-time curve decreased with increasing speed (Table 3, $p<0.0001$), this effect was entirely due to the significant ($p<0.0001$) decrease in trial duration with increasing speed. When the area under the force-time curve was normalized to trial duration this speed effect disappeared. Men tended to produce greater forces, but this effect only reached significance for the area under the force-position curve (43% greater; $p<0.002$). Peak force (20% greater; $p=0.05$) and areas under the corrected force-time curve (21% greater; $p=0.06$) were not significant.

3.4. Effects of direction of footplate motion (OUT vs. IN) on leg press forces (Table 3)

IN overloading of peak forces was not significantly affected by age group, sex, or knee rotation speed. However, older participants – particularly men – did exhibit a weak trend ($p=0.08$) towards greater IN overloading of peak forces (younger men=13%, younger women=24%, older men and women both=27%). IN overloading for areas under the force-time ($p<0.0001$) and force-position ($p<0.0001$) curves were significantly greater for faster speeds, but this effect was not present for peak forces (Figure 3). Women trended towards greater IN overloading for areas under the normalized force-time ($p=0.03$) and force-position curves ($p=0.02$), but these effects also did not reach the threshold for significance. Upon noticing that stronger participants tended to have reduced IN overloading (Figure 4), an additional follow-up analysis was run to test the post-hoc hypothesis that IN overloading of peak forces was correlated with OUT peak force independent of age group and sex. Adding the single best peak OUT force for each participant (as speed effect was not significant) to a linear regression model including age group and sex found that only peak OUT force correlated with IN overloading of peak force ($p=0.01$) and explained 22% more of the variance in the data than the model with only age group and sex (Figure 4).

4. DISCUSSION

4.1. Summary

These multi-joint leg press strength findings largely reflect single-joint strength findings for age (younger were stronger) and direction of footplate motion (IN forces were greater than OUT). Faster knee rotation speeds resulted in significantly greater IN overloading of the areas under the force-time and force-position curves, but not of the peak forces. Women trended towards lower forces, only reaching significance for areas under the force-position curve. Women and older participants trended towards greater IN overloading, but this may have been due to the inverse correlation between peak OUT force and IN overloading of peak forces that was independent of age group or sex. Participant characteristics and clinical test data (Table 1) indicated that the tested population of older adults was very high functioning, thus these findings may not generalize to more impaired populations.

4.2. Hypotheses

Our first hypothesis (younger adults, men, and IN phase would produce greater forces) was supported by younger participants exceeding the forces generated by older participants by 57% for peak forces, 50% for areas under the force-time curves, and 42% for areas under the force-position curves (Figure 2 and Table 3). IN forces were also confirmed to exceed OUT forces by 20% for peak forces and 37% for the areas under both the force-time and force-position curves.

Our peak force data are consistent with the literature for knee extensor testing utilizing similar shortening and lengthening muscle action assessments where younger participants produced 58% greater forces than older participants (Porter et al., 1995). However, we are unaware of any specific analysis of the area under maximal force traces (i.e. work or impulse) for both shortening and lengthening muscle actions.

Men produced greater non-normalized forces here ($p=0.004$), but this effect became non-significant ($p=0.05$) when normalized to body weight. This reduction in sex-related effects when body weight is considered is consistent with the literature, where the percentage difference in force produced by men relative to women drops 23–52% if forces are normalized to body weight (Lindle et al., 1997; Metter et al., 1997; Jozsi et al., 1999). Considering the lack of sex-related decreases in functional mobility despite significant reductions in both strength and weight in women (Reid et al., 2008), strength relative to body weight may be the true metric of interest to assess functional mobility.

Our second hypothesis (OUT forces would decrease and IN forces would increase with faster speeds) was partially supported, as faster knee rotation speeds resulted in greater IN overloading only for the areas under the force-time and force position curves. The lack of any speed effect on IN overloading of peak forces may indicate that peak force data alone could be insufficient to completely assess strength across the ROM. These findings are consistent with previously-reported difficulties in demonstrating these in-vitro effects in-vivo (Enoka, 1996), but speed effects for IN overloading of peak forces might also only be detectable for faster knee rotation speeds (i.e. 100 or 200°/s) and greater differences between tested speeds. As the footplate of the leg press used was not capable of moving much faster

than the maximum speed of 35°/s tested here, we cannot rule out the possibility of speed effects on IN overloading of peak forces under untested conditions. Our third hypothesis (IN forces would be better preserved than OUT in older adults) was rejected, as IN overloading did not significantly differ by age group for peak forces (younger = 18.6%, older = 26.6%), areas under the force-time curves (younger = 25.2%, older = 25.5%), and areas under the force-position curves (younger = 36.1%, older = 41.4%). However, older participants trended towards greater IN overloading, primarily due to the reduced IN overloading of younger males, and significance might have been achieved for a greater sample size. As IN overloading of peak forces was found to be inversely correlated with peak OUT force independent of age group or sex, we suspect that the high level of function and conditioning in our older participants masked the established preservation of lengthening strength with age in less fit populations (Raj et al., 2010; Roig et al., 2010) and that these earlier results may reflect an age-related decline in peak force rather than an effect of age alone. We also found significantly greater IN overloading of the areas under the force-time and force-position curves for faster speeds and trends of greater IN overloading of force in women (Figure 3), but this may also have been due to greater IN overloading in weaker participants.

4.3. Setting Training Targets

The data presented here provide guidance for the use and development of this BLAST leg press used to collect these data, but could also be used to help develop or improve similar motor-driven, closed-chain dynamometer systems. When this device is used for training, the user typically first defines their personal maximum force-position curve using a single maximal OUT trial. The user would then train by keeping their force trace within a specified “window” as the footplate moves OUT and IN at a preset speed. The center of the target window is set at a percentage of the personal maximum curve (a higher percentage is typically used for the IN phase than for the OUT) and the width of this target window is set as a percentage of the force. The footplate starts moving OUT or IN when sufficient force is applied to the footplate to cause the force trace (real-time feedback of force on footplate) to rise into the target window. The force trace then begins to move to the right (for OUT) or left (for IN) and the user must then keep the force trace within the window by pushing on the footplate to control the vertical position of the force trace.

As a subject fatigues, the user often experiences difficulty reaching the bottom of their target window to initiate the IN phase. The different shape of the maximal OUT and IN force-position curves recorded here (Figure 1) provide an explanation for this difficulty- IN forces are greater than OUT, but the peak IN force does not occur until the footplate begins moving back towards the user, so the higher force targets typically used for the IN phase are more difficult to reach while pushing isometrically on a non-moving footplate. To eliminate this problem the target curves for the OUT and IN phases should better represent the actual phase-specific maximum force-position curves. The ideal way to accomplish this would be to define maximal force-position curves for both the OUT and IN phases. An alternate solution that requires fewer setup trials and effort would be scaling up the peak IN force target band by the percentage of IN overload for similar groups (i.e. 13% for younger men, 24% for younger women, and 27% for all unimpaired older adults), then transforming the shape of the IN target curve to more closely correspond to the actual shape of the maximum

IN curve. However, this method would currently not work for populations for which IN overload percentage data do not yet exist, such as the weaker and more impaired populations that training on this device could potentially most benefit.

These data also provide guidance on training speeds, as slower footplate speeds result in greater time spent with muscles active and producing force, while faster speeds allow greater IN overloading of total area under the force-time and force-position curves. It is unknown if these effects may be beneficial to muscle hypertrophy and strength improvement, but considering training specificity, any training regimen to improve functional abilities should ideally incorporate a range of speeds to maximize transfer of strength improvements to functional activities.

4.4. Implications for Training of Older Adults

The literature indicates that training incorporating greater forces on muscle fibers while they are lengthening than they can maintain while shortening can result in greater strength gains that require comparatively less effort and perceived exertion than conventional training (Abbott et al., 1952; Lastayo et al., 1999; LaStayo et al., 2003) and that exercise overloads lengthening strength is safe and effective in older adults (Nichols et al., 1995; Connelly and Vandervoort, 2000; Valour et al., 2004). Delayed-onset muscle soreness can occur for strenuous exercise including lengthening overloading (Miles and Clarkson, 1994) and may be a potential concern, but it is no worse and may even be less severe for older individuals (Clarkson and Dedrick, 1988; Lavender and Nosaka, 2006; Chapman et al., 2008). Our data indicated that weaker persons appear to be capable of greater IN phase overloading (which overloads uniarticular and possibly biarticular muscle fibers while they are lengthening) and thus may attain greater relative strength gains from training using this device. However, the population tested here was quite healthy and strong, so additional data from weaker and more impaired subjects are required to confirm this effect. Overloading the lengthening phase during strength training has been underutilized due to logistical and technical difficulties (Hortobágyi et al., 2001), but the availability of the device tested here (BLAST System; Bio-Logic Engineering Inc., Dexter, MI) and others similar to it (e.g. Exerbotics LLC, Tulsa, OK; Eccentron, BTE Technologies, Inc., Hanover, MD) makes strength training incorporating lengthening overloading more accessible.

Leg press strength training of active, healthy older adults that incorporates IN phase overloading appears to be a safe and promising training method. Leg press training is particularly suitable for training older adults because it is a multi-joint exercise that utilizes biarticular muscles in similar patterns as during functional activities (i.e. supporting body weight while standing, sitting on or standing from a chair) and balance/fall recovery (Pijnappels, Reeves, et al., 2008). Closed kinetic chain exercises like the leg press are also thought to be beneficial for enhancing joint stability due to significant hamstring coactivation (Kvist and Gillquist, 2001). Furthermore, leg press strength is a strong independent predictor of the ability to successfully recover from a lab-induced trip (86% sensitivity and 90% specificity (Pijnappels, van der Burg, et al., 2008)) and the transfer of strength improvements from leg press training to balance recovery have been shown to be feasible (Pijnappels, Reeves, et al., 2008).

4.5. Limitations

This study was unpowered, as no published data exist on this device. As such, it may have been underpowered to detect possible effects of knee rotation speed on forces and age group on IN overloading. The high level of function and strength in our older participants may also not reflect the older population as a whole. This device is not capable of testing speeds much faster than about 35°/s (exact maximum speed depends on user height), which may have also played a role in speed effects on IN overloading of peak forces not reaching significance. This limitation on footplate actuation speed means that this leg press also cannot train at high speeds to preferentially train muscle power, which may be more beneficial for improvements in physical function (Porter, 2006). However, the speed range of this device is within that commonly used during loaded functional activities. For example, a healthy young subject stands from a chair in 1.6 ± 0.1 s and sits in 1.6 ± 0.2 s (Highsmith et al., 2011), while these 35°/s trials were completed in only 1.30 ± 0.19 s for the OUT phase and 1.35 ± 0.19 s for the IN phase.

A potential limitation of leg press training is that it may not effectively train plantarflexor strength. The age-related loss of plantarflexor strength may be partially responsible for the age-related redistribution of joint torques and impairments of gait (Kerrigan et al., 1998; DeVita and Hortobagyi, 2000; Silder et al., 2008) and is most severe in fallers (Whipple et al., 1987). Considering this, a plantarflexor training component should be incorporated in any strength training program designed for older adults. We have developed a potential method of plantarflexor training that can be administered using this device or any other leg press that consists of a block of wood under the forefeet (high enough to prevent the heels from contacting the footplate) accompanied by an increase the maximum footplate displacement. These small modifications place a high demand on the plantarflexors by requiring all footplate forces to be applied by the forefeet and allow the ankles to traverse a greater ROM.

5. CONCLUSIONS

Leg press strength training on a closed-chain dynamometer is a promising new training method that appears to have the greatest potential benefits for the weakest participants. Training target profiles should be set based on participant-specific maximums across the ROM in both OUT and IN directions to avoid undertraining portions of the ROM, account for participant differences in IN overloading, and prevent difficulties with initiating the IN phase. Faster knee rotation speeds did not affect peak forces, but did increase the areas under the force-time and force-position curves, so target profiles should be based on data recorded at a speed in the middle of the range to be trained in order to best generalize to all training speeds.

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Abbreviations

ROM Range of motion

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Figure 1.

Leg press force data of a single maximal effort trial in each direction for a representative participant. Arrows along curves indicate direction of footplate motion during OUT and IN phases of trial.

Figure 2.

Leg press peak force data in percentage of body weight plotted by age group (columns), sex (rows), knee rotation speed (x-axis), and footplate direction. Effects of age group and footplate direction significant ($p < 0.0001$).

Figure 3.

Mean (standard deviation) of leg press force IN overloading, defined here as the percentage improvement of the IN value relative to the OUT value, plotted by sex and knee rotation speed. IN overloading of peak force (left) was unaffected by knee rotation speed, but the areas under the force-time curve (center) and under the force-position curve (right) both significantly increased with knee rotation speed ($p < 0.0001$). Women tended to exhibit greater IN overloading for the area measures ($p = 0.03$ for center and $p = 0.02$ for right), but none of these effects reached the corrected level of significance ($p < 0.002$). Note that the areas under the force-time curve (center) were corrected for trial duration to account for the significantly shorter ($p < 0.0001$) trials at faster speeds.

Figure 4.

Scatter plot of the single best value for IN overloading of peak force (improvement of peak IN phase force relative to peak OUT phase force) against single best leg press peak force across all knee rotation speeds (no speed effect found) during OUT phase for each participant. Significant ($p=0.01$) inverse correlation was found between peak OUT force and IN overloading that was independent of age group and sex.

Table 1

Mean (standard deviation) of participant characteristics and clinical test results.

	Younger (<i>n</i> =14)		Older (<i>n</i> =14)	
	Men (<i>n</i> =7)	Women (<i>n</i> =7)	Men (<i>n</i> =7)	Women (<i>n</i> =7)
Age (years)	23.7 (2.8)	29.4 (4.3)	72.9 (6.7)	72.0 (3.8)
Age Range	20–29	23–35	66–82	68–79
Height (cm) ^a	180.9 (7.7)	165.3 (9.5)	179.9 (5.2)	160.9 (6.7)
Weight (kg) ^{a,c}	82.5 (10.2)	60.2 (8.4)	86.2 (8.5)	71.2 (4.7)
Body Mass Index (kg/m ²) ^c	25.2 (2.3)	22.1 (3.5)	26.6 (1.8)	27.4 (2.7)
Gait Speed- Preferred (m/s)	1.29 (0.14)	1.31 (0.17)	1.37 (0.20)	1.33 (0.15)
Gait Speed- Fast (m/s)	2.32 (0.44)	2.23 (0.35)	2.38 (0.33)	1.96 (0.21)
Functional Reach (cm) ^c	48.6 (6.3)	42.5 (8.1)	41.0 (5.6)	39.6 (4.8)
Maximum Step Length (cm) ^{b,e}	136.7 (10.4)	117.9 (16.4)	115.3 (7.4)	91.7 (15.5)
Maximum Step Length (%ht) ^e	75.6% (6.0%)	70.9% (7.0%)	64.2% (5.1%)	56.8% (8.7%)
Timed Up and Go (s) ^c	4.58 (0.39)	4.29 (0.46)	4.97 (1.19)	5.18 (0.51)
Four Square Step Test (s) ^d	6.06 (1.69)	5.63 (1.09)	7.08 (1.51)	7.43 (0.91)

^a $p < 0.0001$ sex;

^b $p < 0.01$ sex;

^c $p < 0.05$ age group;

^d $p < 0.01$ age group;

^e $p < 0.001$ age group

Table 2

Improvement in R^2 value of correlations between functional clinical tests and leg press force measures due to normalizing forces to body weight. Force data used are maximum across all speeds tested.

	Footplate Moving OUT			Footplate Moving IN			MEAN
	Peak Force	Area Under Force-Time Curve	Area Under Force-Position Curve	Peak Force	Area under Force-Time Curve	Area Under Force-Position Curve	
Maximum Step Length (%ht)	0.117	0.109	0.167	0.136	0.103	0.194	0.138
Timed Up and Go (s)	0.058	0.054	0.028	0.079	0.064	0.039	0.054
Four Square Step Test (s)	0.051	0.036	0.036	0.060	0.038	0.047	0.045
MEAN	0.075	0.067	0.077	0.092	0.068	0.093	0.079

Table 3

Mean (standard deviation) of leg press force data by age group and sex.

	Younger (<i>n</i> = 14)		Older (<i>n</i> = 14)	
	Men (<i>n</i> = 7)	Women (<i>n</i> = 7)	Men (<i>n</i> = 7)	Women (<i>n</i> = 7)
Peak Forces (N) ^{a,b}				
Footplate moving OUT	4,793 (1,443)	2,770 (360)	3,033 (1,083)	2,021 (348)
Footplate moving IN	5,417 (1,699)	3,414 (588)	3,791 (1,204)	2,532 (468)
Peak Forces Normalized to Body Weight (% wt) ^{b,c}				
Footplate moving OUT	588% (129%)	475% (75%)	364% (140%)	290% (49%)
Footplate moving IN	665% (160%)	583% (101%)	454% (154%)	364% (66%)
Area Under Normalized Force-Time Curves (% wt*s/s) ^{b,c}				
Footplate moving OUT	3.74 (0.91)	2.96 (0.49)	2.44 (0.79)	1.97 (0.37)
Footplate moving IN	4.90 (0.82)	4.16 (0.82)	3.30 (1.09)	2.83 (0.50)
Area Under Normalized Force-Position Curves (% wt*m) ^{a,b}				
Footplate moving OUT	0.590 (0.133)	0.389 (0.092)	0.393 (0.132)	0.276 (0.040)
Footplate moving IN	0.776 (0.179)	0.539 (0.119)	0.536 (0.176)	0.405 (0.066)

These data are intersubject means across the best trial for each subject at all three speeds.

Area under force-time curve normalized to trial duration to account for significantly ($p < 0.0001$) briefer trials for faster speeds.

^a $p < 0.002$ sex;

^b $p < 0.0001$ footplate direction;

^c $p < 0.002$ age group