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Effect of robotic performance-based error-augmentation versus error-reduction training on the gait of healthy individuals

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Abstract

Effective locomotion training with robotic exoskeletons requires identification of optimal control algorithms to better facilitate motor learning. Two commonly employed training protocols emphasize use of training stimuli that either augment or reduce performance errors. The current study sought to identify which of these training strategies promotes better short-term modification of a typical gait pattern in healthy individuals as a framework for future application to neurologically impaired individuals. Ten subjects were assigned to each of a performance-based error-augmentation or error-reduction training group. All subjects completed a 45-min session of treadmill walking at their preferred speed with a robotic exoskeleton. Target templates prescribed an ankle path for training that corresponded to an increased step height. When subjects' instantaneous ankle positions fell below the inferior virtual wall of the target ankle path, robotic forces were applied that either decreased (error-reduction) or increased (error-augmentation) the deviation from the target path. When the force field was turned on, both groups walked with ankle paths better approximating the target template compared to baseline. When the force field was removed unexpectedly during catch and post-training trials, only the error-augmentation group maintained an ankle path close to the target ankle path. Further investigation is required to determine if a similar training advantage is provided for neurologically impaired individuals.

Keywords

gait; robotic exoskeleton; force field; rehabilitation; locomotion

Introduction

Robotic lower limb exoskeletons have the potential to assist gait rehabilitation for individuals with neurological dysfunction. Robotic devices allow for more intensive and

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repetitive training than conventional physical therapies and as such, may enhance motor plasticity [1-3]. Although some positive outcomes have been demonstrated [4-5], the effectiveness of robotic-assisted gait training is still controversial [6-8].

It is still unclear how different force control strategies implemented on robotic devices affect the process of motor learning or promote neural recovery [9-10]. The principle of assist-as-needed derived from the clinical convention has been implemented in the control algorithms of some current robotic devices [11-13]. The assistance is used to reduce subject's performance error relative to the prescribed behavior. We refer to this paradigm here as error-reduction. The major advantage of error-reduction strategy is that correct movement patterns can be guided, but without forcing the movements through one fixed path [14]. However, this strategy tends to minimize movement errors [12], and subjects' knowledge of performance error is important for motor learning [15-17]. It also has been demonstrated that the human motor system attempts to decrease levels of muscle recruitment (i.e., slacking) when movement errors are reduced during performance of a dynamic task [18-20]. Thus, an error-reduction strategy may substantially reduce patients' effort.

There have been a few examples of applying robotic resistance in training functional tasks [9, 21-24]. To date, the majority of these studies have applied a fixed, velocity-dependent force-field that amplifies movement error independent of subjects' on-line performance. The after-effect seen upon removal of the force field has been proposed as necessary to train correct movement patterns following neurological dysfunction [23-24]. However, after-effects have been shown to be short lasting after removing a force field [17, 25-26]. Thus, in addition to the possible strengthening effect from training with the robotic resistance, benefits occurring from after-effects need to be further determined.

An alternative control strategy is to apply a performance-based resistance that amplifies error based on subjects' on-line performance. Performance-based resistance has the potential to provide additional kinetic and proprioceptive biofeedback (e.g., increased mechanical work) to subjects when their gait patterns deviate from a desired pattern. One example of robotic performance-based resistance was developed by Simon et al. [21]. During a leg extension task, a resistive load was applied against leg extension, with the amount of resistive load proportional to the difference of force generation between the two legs. With performance-based resistive force cues, healthy subjects altered their lower limb forces toward a target force level and were able to reproduce the target level when the force cues were removed. This finding indicates that a performance-based resistance may be effective to shape the motor outputs for a dynamic task, and may persist when the force cuing is not present.

The purpose of the current study was to investigate whether performance-based robotic training using an error-augmentation algorithm better facilitates short-term changes of a typical gait pattern in healthy individuals compared to robotic training employing an error-reduction algorithm. Healthy individuals were studied as a starting point, serving as a proof-of-concept for future work with neurologically impaired individuals. We hypothesized that error-augmentation gait training would lead to walking post-training with ankle paths closer to a prescribed path that was altered from normal within a single training session compared to the performance of subjects receiving error-reduction gait training.

Methods

Subjects

Twenty healthy subjects gave written informed consent, approved by the university's Institutional Review Board, to participate in this study. Volunteers were stratified by gender and randomly assigned to either the error-reduction (5 females, 5 males; age: 21.8 ± 2.8 years; height: 172.2 ± 7.3 cm; mass: 62.1 ± 7.2 kg) or error-augmentation (5 females, 5 males; age: 20.8 ± 0.9 years; height: 170.4 ± 7.3 cm; mass: 67.3 ± 12.7 kg) group.

Device description

Force fields were applied to subjects' right leg using a robotic leg exoskeleton (Figure 1). Details of the exoskeleton's design are documented elsewhere [11, 27-28]. The force-field controller was developed to apply tangential and normal forces at subjects' ankle [11, 27]. Linear actuators mounted at the hip and knee joints of the exoskeleton provided a pattern of torques simulating the desired forces applied to the ankle.

Target walking templates based on the spatial paths of individual subject's ankle locations (i.e., ankle path) were created for training (Figure 2). Target templates required subjects to step 57% or at least 3.4-cm higher than their baseline step height. This new walking template was chosen, based on pilot experiments, to provide sufficient challenge to the subjects while at the same time minimizing fatigue. Many individuals with neurological impairments have insufficient hip and knee flexion during swing, and one goal of walking training is to increase their step height to minimize the likelihood of tripping. Thus, we chose to use a template that required a greater as compared to a shallower step height than normal in this proof-of-concept study so that the required change in walking pattern had similarities to what would be expected in neurologically impaired individuals and, therefore, would be more relevant to future applications to that population.

The force-field controller was modified so that normal forces were provided only when subject's ankle positions were below the inferior virtual wall of the prescribed path (i.e., unidirectional) during swing (Figure 2). When subjects' instantaneous ankle positions fell below this wall, the error-reduction group received normal forces tending to bring their ankle positions towards the target position. For the error-augmentation group, subjects received normal forces tending to take their ankle positions further away from the target. The error here was defined as a deviation from the inferior virtual wall about the target template. The amount of robotic resistance or assistance varied depending on the amount of deviation from the prescribed paths (i.e., subjects received a spring-like force).

Experimental design and protocol

Subjects wore the exoskeleton while walking on the treadmill at their preferred speed (1.4 ± 0.1 mph). They were given 10 minutes of familiarization time, after which they walked in the exoskeleton without the force field for 10 minutes (Baseline). They then walked with the force field in nine, 5-minute training bouts (T1-T9). This was followed by two, 5-min post-training bouts (P1-P2) without the force field. Ten trials of over-ground walking were evaluated immediately following baseline and P1. Four 30-sec catch-trials were administered immediately after completing training bouts T1, T3, T5, and T7.

Subjects were informed that a target gait pattern was established and that the amount of robotic force they could receive would vary based on the amount of deviation from the target pattern. Their task was to discover the target pattern by minimizing those forces. Subjects were not informed explicitly about the target template, the type of robotic force (error-reduction or augmentation), or removal of the force field during catch and post-training trials.

Data acquisition and analysis

Right lower-limb joint kinematics and bilateral foot-switch data were collected while walking with the leg exoskeleton. For the over-ground walking, the kinematic data were collected with an 8-camera Qualisys (Gothenburg, Sweden) motion capture system (120-Hz). The deviation between subjects' actual and prescribed ankle paths during treadmill walking were estimated by the area enclosed between the two paths during swing (Total Area) for every 30-second trial. A smaller total area indicates less deviation from the prescribed paths. We also identified and summed regions where the actual ankle paths were above (Area Above) or below (Area Below) the prescribed path. Because the sizes of the ankle paths varied among subjects, each subject's data was normalized to the deviated area between their average baseline and the target ankle path. To evaluate changes in over-ground walking, we estimated (1) the area enclosed between the ankle paths obtained during the swing phase of the pre- and post-training sessions, and (2) changes of step height between pre- and post-training.

Statistics

Repeated measures analyses of variance (RM-ANOVAs) were used to test for differences between groups and across testing times for the deviated areas (Total Area, Area Below and Area Above), and for joint kinematics (peak hip flexion, knee flexion, dorsiflexion during swing and peak plantar flexion at pre-swing). The analyses were performed separately for three combinations of testing periods: baseline versus training (4 periods: B, T1, T5, T9), baseline versus catch trials (5 periods: B, CT1-4), and baseline versus post-training (3 periods: B, P1-2). For the over-ground walking data, another RM-ANOVA was used to test for differences in the deviated area of ankle paths and the changes of step height before and after training between legs (trained and untrained) and groups. Statistics were performed in SPSS™ (IBM Corp., Somers, NY) PAWS version 18. The significance level was set at $p<0.05$ and significant interactions investigated with post-hoc tests with Bonferroni corrections.

Results

Area between the actual and prescribed ankle paths

Training bouts—While the force field was turned on, both groups of subjects walked with ankle paths that were close to the prescribed templates, evidenced by the significantly smaller Total Area ($p<0.001$) compared to the baseline (Figure 3). The error-augmentation group gradually reduced the Area Below and increased the Area Above during training (Figure 3A). By the last training bout, the error-augmentation group showed a 33% reduction of Total Area (T9: 0.67 ± 0.20, mean ± SD, $p=0.012$). They also exhibited a reduced Area Below (T9: 0.15 ± 0.17, $p<0.001$) and increased Area Above (T9: 0.51 ± 0.28, $p<0.001$) compared to baseline performance. The error-reduction group (Figure 3B) also had a 37% reduction in Total Area (T9: 0.63 ± 0.15, $p<0.001$), a smaller Area Below (T9: 0.57 ± 0.14, $p<0.001$) and greater Area Above (T9: 0.06 ± 0.05, $p=0.024$) compared to baseline. The error-reduction group had a similar Total Area ($p>0.05$) but greater Area Below ($p<0.001$) and smaller Area Above ($p<0.001$) compared to the error-augmentation group.

Catch-trials—The error-augmentation group retained the new ankle paths when the robotic force was removed unexpectedly whereas the error-reduction group walked with ankle paths similar to their baselines (Figure 3). The error-augmentation group had significantly smaller Total Area (all $p<0.01$), smaller Area Below (all $p<0.001$) and greater Area Above (all $p<0.05$) for all catch-trials compared to the baseline (Figure 3A). In contrast, the error-reduction group showed no effect of testing period on Total Area ($p=0.18$) or Area Below ($p=0.09$) between baseline and the catch-trials (Figure 3B), despite a trend toward reduction. For Area Above, the error-reduction group had significantly greater area during the last two catch-trials ($p<0.05$) compared to the baseline, although the area was quite small. Compared to the error-augmentation group, the error-reduction group had significantly greater Total Area (p=0.0014), greater Area Below (p<0.001), and smaller Area Above (p<0.001).

Post-training bouts—The error-augmentation group consistently walked with ankle paths different from their baseline during post-training. Although the error-reduction group demonstrated a similar effect, it was much smaller (Figure 3). Compared to the baseline, Total Area was significantly smaller for both groups (p<0.001). Compared to the error-augmentation group, the error-reduction group had significantly greater Area Below and smaller Area Above (all p<0.01). Thus, although the error-reduction group showed an improvement by approximating the target template, this modification was much smaller than that exhibited by the error-augmentation group.

Joint kinematics

Changes in subjects' ankle paths required corresponding modifications in the joint kinematics. During training, all subjects walked with increased hip and knee flexion during swing compared to baseline performance (Figure 4). By the last training bout, the error-augmentation group had significantly greater peak hip flexion by ~7 degrees (B: 17.2° ± 2.4°, T9: 24.3° ± 2.5°, p<0.001) and peak knee flexion by ~9 degrees (B: 48.2° ± 4.7°, T9: 57.5° ± 2.8°, p<0.001) during swing. For the error-reduction group, subjects had significantly greater peak hip flexion by ~4 degrees (B: 17.4° ± 4.9°, T9: 21.3° ± 4.3°, p<0.001) and peak knee flexion by ~3 degrees (B: 47.9° ± 10.3°, T9: 51.3° ± 10.3°, p<0.01).

The error-augmentation group walked with greater hip (~7°) and knee (~5°) flexion during the swing phase for most catch-trials compared to baseline performance (all p<0.01 except knee flexion for the first catch-trial) (Figure 5). For the error-reduction group, peak hip and knee flexion during the swing phase were similar to the baseline (all p>0.05).

During post-training trials, the error-augmentation group consistently walked with ~6° greater swing-phase hip flexion (p<0.001) than during baseline performance while the error-reduction group had increased swing-phase hip flexion by 2° only at P2 (p<0.05) (Figure 5). Overall, peak knee flexion during swing was significantly greater compared to the baseline (p<0.001), although there was no significant effect of group (p>0.56) nor an interaction of group with test period (p>0.14).

Although ankle motion was not directly targeted by the training, changes in its motion were investigated as well. There were no significant effects of group, test period or their interaction (all p>0.05) for the peak plantar flexion at pre-swing or peak dorsiflexion during swing.

Over-ground walking trials

Both groups had similar ankle paths and stepping height during over-ground walking before and after the training. The deviated area enclosed between baseline and post-training ankle paths were not significantly different between legs (p>0.5) or from zero (p>0.05) for either group, and there was no group difference (p>0.5). Changes in step height from pre- to post-training did not differ from zero (p>0.05), nor did they differ between legs (p>0.5) or groups (p>0.8).

Discussion

The findings of this study support our hypothesis that error-augmentation force-field training would lead to a greater short-term modification of subjects' step height than training with an error-reduction force field. During training, subjects in both groups walked with ankle paths close to the target paths. However, only those receiving error-augmentation training showed substantial changes when removing the force field. In addition, changes in the ankle path induced by this single-session training were not transferred to the over-ground walking in this group of neurologically intact subjects.

The current findings for error-reduction training differ from those of a previous study by Kim et al. [28]. That study showed a persistent modification of subjects' ankle paths after error-reduction training when the force-field training included visual feedback. In the current study, visual feedback was not provided because our primary interest was to investigate differences in the type of force feedback. Moreover, subject's attempts to reduce their ankle path deviation from the template based on visual error might minimize the amount of force feedback. Kim et al. (2010) demonstrated that subjects receiving combined visual and force-field feedback exhibited a more persistent change in their ankle paths than subjects receiving either type of feedback alone. Whether adding visual feedback in the current study would reduce or enhance the differences between the groups' performance is a question requiring further investigation. Another important difference between Kim et al. (2010) and this study is that those subjects were asked to produce a shallower ankle path than normal. This may have been easier to learn than increasing the step height, which requires more physical effort (i.e., to lift the leg further against gravity).

Movement economy may explain the differences in the areas above and below the target template exhibited in the two training groups. It has been demonstrated that humans preferably choose movement patterns that require minimum physical energy while performing a dynamic task [29-30]. In this study, the error-reduction group walked with ankle paths that fell below the target throughout most of the training while the error-augmentation group walked with ankle paths predominantly above the template during and following training. From the aspect of gait energetics, leg swing consumes ~30% of net energy required for walking [29]. Thus, it is costly to lift the leg higher than required. By stepping lower than the target template, the error-reduction group might have benefited energetically from the spring-like assisting forces that tended to bring their ankle positions toward the target. In contrast, when the error augmentation group deviated further below the target template, they received forces that tended to push them even further away, which might have been even more costly energetically to overcome than if stepping higher than the template, where no external force was experienced. Therefore, stepping a little higher than the target might have been a more economic strategy for the error-augmentation group.

The use of performance-based resistive forces applied by a robot has potential application to gait retraining for neurologically impaired individuals. The current study comparing two training strategies with healthy individuals served as proof-of-concept for future work with neurologically impaired individuals. Although achieving optimal movement economy may not be the priority for an impaired nervous system while performing tasks, the performance-based resistive forces can provide strong proprioceptive and kinetic cues that might work better than a visual or verbal cuing. In addition, given the results of the current study, error-augmentation training might be a better training stimulus than error reduction for persons with neurological disorders [24]. However, this issue requires further investigation because the nature of the response to either training strategy may depend on the severity of a person's motor impairments. For example, error-augmentation training requires some degree of independent stepping. Based on initial studies in our laboratory, at least some stroke survivors with severe motor impairments cannot tolerate the resistive forces provides by error-augmentation training. Error reduction training may be the only option for such individuals. This suggestion is consistent with recent experiments revealing differences in response to similar training paradigms in both healthy subjects and neurologically impaired individuals, depending on their initial skill or impairment level [30-31].

A major limitation of this study is that the force field we tested was unidirectional, applying the force only when subjects' actual ankle positions fell below the prescribed templates. Although the majority of neurologically impaired individuals have demonstrated a much shallower step height than healthy controls, some stroke survivors exhibit an exaggerated step height by compensations such as increased hip flexion and pelvic hiking to compensate for limited knee flexion. Indeed, we are implementing a bi-directional error-augmentation force field to train walking post-stroke. Other limitations of the robotic training approach are the added inertia provided by the robot that trainees must overcome as well as some limitations on leg and pelvic motion by the exoskeleton. These factors could also limit training with robotic forces, such as increasing fatigue. However, fatigue is unlikely a factor affecting the results in the current experiment because only the error-augmentation group received resistive forces and this group had the strongest training effect. Finally, the current setup did not allow obtaining information about kinematics of the non-trained leg, which might have provided useful additional information for interpreting the results of the two training strategies.

Conclusion

Neurologically intact subjects were able to walk with stepping patterns closer to a prescribed template that required a higher than normal step height. Matching the target template was substantially better in persons receiving error-augmenting forces compared to error-reducing forces. Future studies will examine performance-based robotic force fields in neurologically impaired populations.

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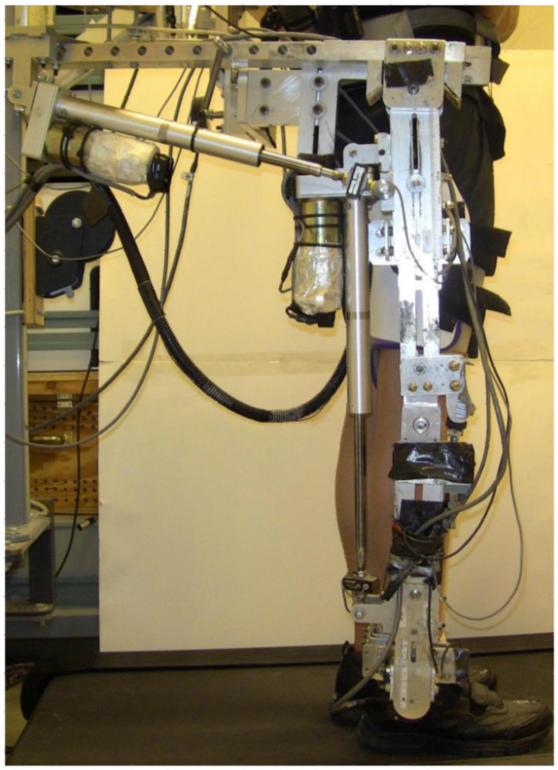

Figure 1.

Subjects wore the robotic leg exoskeleton on their right lower limb. The unilateral robotic leg exoskeleton has four segments, i.e., pelvis, thigh, shank and foot that can be adjusted to each subject's stature. The exoskeleton design allows for sagittal plane movements (e.g., flexion/extension) of the hip, knee and ankle, frontal plane movements (i.e., adduction/abduction) of the hip, and trunk rotation. The linear actuators at the hip and knee joints power the sagittal plane joint movements.

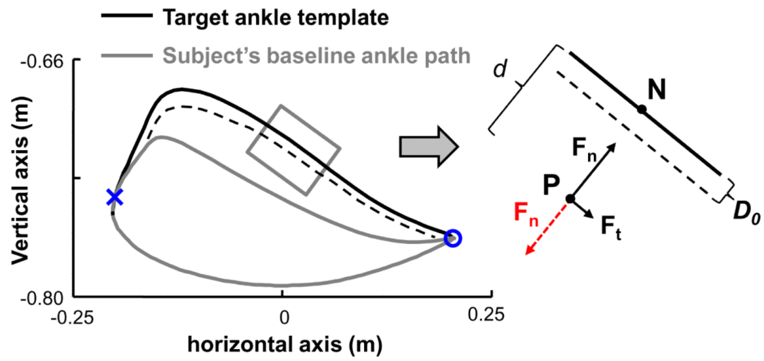

Figure 2.
Example ankle path template with origin set at the hip joint. The blue circle indicates the timing of heel strike and the blue cross indicates the timing of toe-off. The target ankle template was created individually for subjects based on their baseline ankle path and step height. The distance '*d*' is defined from the current ankle position (P) to the nearest point on the prescribed ankle path (N). D_0 is defined as the width of virtual wall (in millimeters). The normal forces (F_n) were produced only when the distance between P and N exceeded D_0. The tangential forces (F_t) were minimal, designed to ensure that the subjects' leg produced continuous movement along the ankle path. When subjects' instantaneous ankle position (P) fell below the virtual wall (the black dashed line) of the nearest point on the target template (N), the performance-based error-augmentation algorithm led to generation of a spring-like force of positive stiffness by the motors, tending to take the subject's leg further away from the target (N) (see the F_n, dashed red line). In contrast, the error-reduction algorithm led to a spring-like force being generated with negative stiffness, tending to bring subject's foot towards the target (N) (see the F_n in black line). The amplitude of the normal force (F_n) was proportional to the deviation between subjects' instantaneous ankle position and the prescribed ankle position.

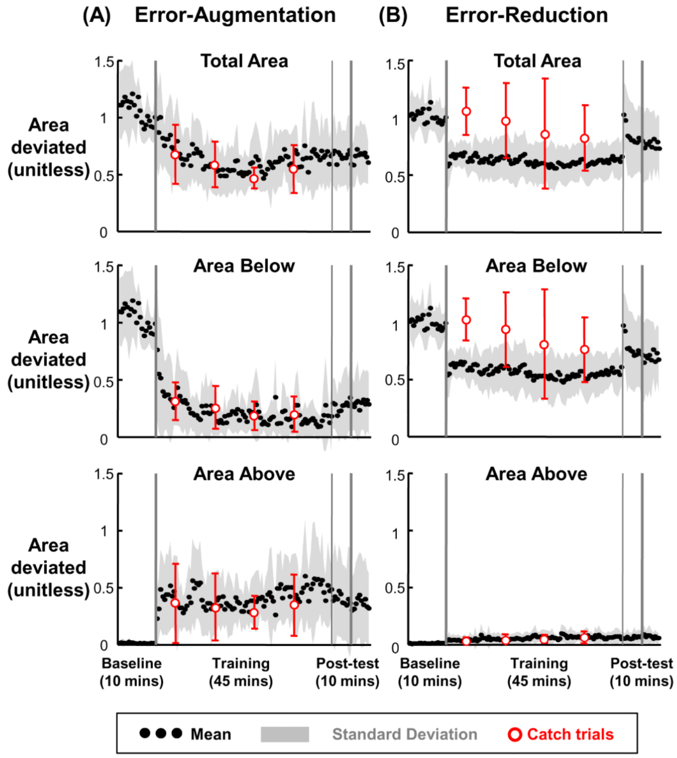

Figure 3.
Average ±1 SD (gray area) of the normalized area between the actual ankle path and the prescribed ankle path is presented. Average data (filled black dots) are shown for each 30-second trial during baseline walking, training epochs and post-training walking. The open red circles and associated error bars represent the average ± 1 SD from catch trials.

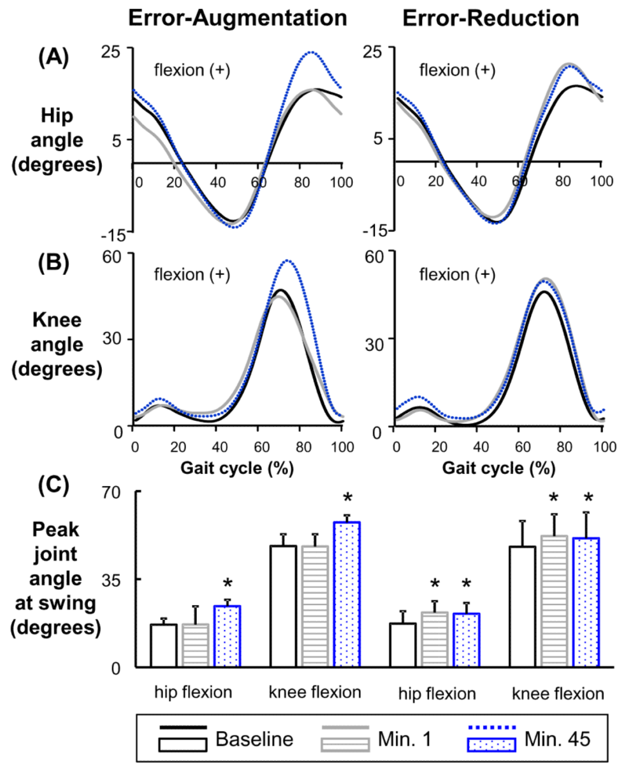

Figure 4.

(A) Hip, (B) knee joint angle profiles and (C) peak swing-phase joint angles are shown for last minute of baseline (baseline, solid black), the first minute of force-field training (minute 1, solid grey), and the last minute of force-field training (minute 45, dotted blue). Joint angle profiles were time normalized to gait cycles, from heel strike to heel strike. Data are the average of all subjects in each group. Error bars indicate standard deviation. *Significant differences from the baseline are labeled for the time periods. During training, all subjects walked with increased hip and knee flexion during the swing phase compared to the baseline walking.

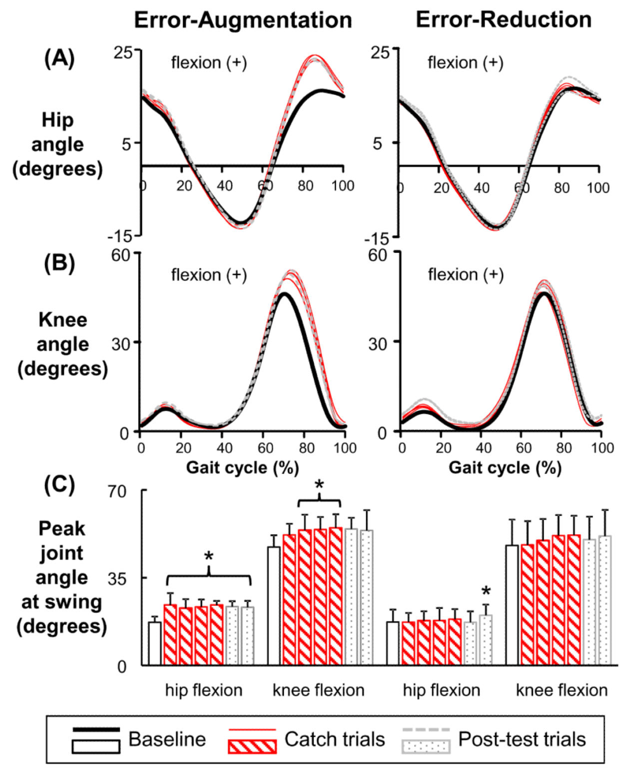

Figure 5.

(A) Hip, (B) knee joint angle profiles and (C) peak swing-phase joint angles are shown for last minute of baseline (baseline, solid black), during four catch trials (catch trials, solid red), the first and last minutes of post-training trials (post-test trials, dashed grey). Joint angle profiles were time normalized to gait cycles, from heel strike to heel strike. Data are the average of all subjects in each group. Error bars indicate standard deviation. *Significant differences from the baseline are labeled for the time periods. The error-augmentation group consistently walked with increase hip flexion at the swing phase during catch and post-training trials. However, the error-reduction group showed an increase in hip flexion only at the second post-training trial. For knee joint kinematics, the error-augmentation group had greater knee flexion at the swing phase during catch trials while the error-reduction group had similar knee joint angles compared to the baseline.