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Robotics for Stroke Recovery

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Abstract

The introduction of robotics into neurorehabilitation is a relatively recent phenomenon. To date, both their acceptance by the rehabilitation community and penetration of robotic devices into rehabilitation facilities have been limited. The majority of clinical studies evaluating the efficacy of rehabilitation robotics to date have framed the question in terms of superiority between robotic approaches and some chosen standard therapy. Not surprisingly, the results of many of these studies have revealed nonsignificant differences between robotic and traditional rehabilitation approaches, which clinicians generally interpret as failure of the robotic approach. Improvements in response to both traditional and robotic approaches yielding results of "no difference" could, however, be interpreted as a positive result. Considered in this light, robotic approaches may offer the opportunity for therapeutic intervention to more individuals and/or may extend the therapeutic opportunity (i.e., dose, time, repetitions) while practitioners focus on other critical aspects of rehabilitation. Here, it is

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important to recognize that many of the robotic approaches have been designed to mimic currently utilized rehabilitation interventions, at least as these interventions are currently understood. Thus, the modest efficacy of robotic approaches demonstrated to date may stem from limitations in our current understanding of the critical processes of neural recovery, and how to effectively induce neural recovery, rather than from limitations of robotic devices per se. This chapter considers the problem of walking dysfunction following stroke and offers perspectives on the use of the Lokomat to promote walking recovery.

Keywords
Biomechanics • EMG • Locomotion • Neurorehabilitation • Recovery • Sensorimotor integration • Stroke

15.1 Introduction

The introduction of robotics into neurorehabilitation is a relatively recent phenomenon. To date, both their acceptance by the rehabilitation community and penetration of robotic devices into rehabilitation facilities have been limited. These circumstances may reflect the state of rehabilitation practice, in general, rather than limitations of robotic approaches for rehabilitation. Among clinical practitioners, the reaction to rehabilitation robotics tends to be bimodal: either they are embraced for the potential opportunity to offer more therapy, more consistent therapy and perform the numerous repetitions required to induce neural plasticity; or they are met with suspicion and criticized for their limitations, expense, and attendant technical challenges. Indeed, at least in North America, many practitioners argue for the superiority of traditional, "hands-on" therapeutic approaches [1].

Of note, the majority of clinical studies evaluating the efficacy of rehabilitation robotics to date have framed the question in terms of superiority between robotic approaches and some chosen "standard" therapy. Perhaps not surprisingly, the results of many of these studies have revealed nonsignificant differences between robotic and traditional rehabilitation approaches [1–3]. Clinicians generally interpret these nonsignificant differences as failure of the robotic approach. Importantly, however, improvements in response to both traditional and robotic approaches yielding results of "no difference" could be interpreted as a positive result. Rehabilitation produces measurable effects! Considered in this light, robotic approaches may offer the opportunity for therapeutic intervention to more individuals and/or may extend the therapeutic opportunity (i.e., dose, time, repetitions) while practitioners focus on other critical aspects of rehabilitation. Here, it is important to recognize that many of the robotic approaches have been designed to mimic currently utilized rehabilitation interventions, at least as these interventions are currently understood. Thus, the modest efficacy of robotic approaches demonstrated to date may stem from limitations in our current understanding of the critical processes of neural recovery, and how to effectively induce neural recovery, rather than from limitations of robotic devices per se. This chapter considers the problem of walking dysfunction following stroke and offers perspectives on the use of the Lokomat (Hocoma, Volketswil, Switzerland) to promote walking recovery.

15.2 Statement of the Problem

Stroke is the leading cause of serious, chronic disability in adults worldwide. Each year, approximately 16 million strokes occur worldwide. As a result of marked improvements in acute stroke management, the cohort of survivors now exceeds 62 million persons worldwide, a third of whom experience significant physical disability and functional impairment [4]. Because its risk doubles with each decade of age beyond 55 years, stroke has historically been considered a problem of an aging population. However, in the last decade, the representative demographics have shifted dramatically to reveal an emerging representation of younger individuals affected by stroke. Approximately half of total stroke costs [5] are now directed toward persons between the ages of 45 and 64 years. This demographic shift heralds an urgent and critical need to improve the efficacy and effectiveness of stroke rehabilitation. This need encompasses strategies not only to reduce stroke-related costs and disability but also to restore function for persons in the productive and vital years of their lives. Simply put, we need to increase our expectations of the outcomes attainable in stroke rehabilitation.

Walking dysfunction is one of the greatest physical limitations contributing to stroke-related disability. While two-thirds of persons who suffer a stroke regain ambulatory function, their gait is slow, asymmetrical, and metabolically inefficient [6–9]. These characteristics are associated with difficulty advancing and bearing weight through the more affected limb, leading to instability and an increased risk of falls [10]. Secondary impairments, including muscle disuse and reduced cardiorespiratory capacity, often contribute to further declines in gait function. Walking dysfunction restricts independent mobility, and autonomy, and therefore severely impacts quality of life for many stroke survivors and their families [11, 12]. Given this constellation of problems, it is not surprising that improved walking is one of the most frequently articulated goals of neurorehabilitation [13]. Interventions that effectively restore and promote meaningful recovery of walking function are needed to enable these persons to resume participation in their premorbid social roles. This challenge offers a significant opportunity for the area of rehabilitation robotics.

15.3 Walking Recovery Poststroke

Traditional approaches to gait therapy involve one-on-one treatment by a physical therapist using various forms of exercise, equipment, and feedback [14, 15]. However, because many hemiparetic persons are unable to bear weight normally through the paretic limb and lack normal tolerance to upright posture, such traditional gait training fails to establish the requisite biomechanical conditions for normal locomotion. Indeed, this traditional, clinical approach may engender, and even reinforce, compensatory movement strategies and dependence on assistive devices. Concern regarding traditional gait training approaches extends beyond compromised biomechanics to the sensory-perceptual aspects of locomotor control. Integration of inaccurate and inappropriate sensory information is disruptive and can interfere with positive effects of motor rehabilitation, especially at critical stages in the process of neural recovery [16].

Recent efforts have emphasized the need for a "task-specific" [17] approach to gait therapy based on fundamental concepts of motor learning [18] and specificity of training [19]. Indeed, the task-specific approach appears to produce somewhat greater gains in gait function than traditional therapy [20]. Of note, results reported by proponents of the task-specific approach reveal that persons in the subacute period poststroke demonstrating at least a minimal level of gait function pretreatment (i.e., ability to walk at ~0.3 m/s) are most likely to produce significant treatment-induced improvements in walking [21]. This observation suggests that baseline hemiparetic severity may be the fundamental determinant of locomotor outcome; that is, a critical level of function must be retained following stroke to benefit from gait rehabilitation. This perspective suggests a limited capacity for locomotor recovery in persons poststroke.

An emerging, contemporary approach to walking recovery involves partial body weight-supported treadmill training, also termed "Laufband" therapy [22, 23] or locomotor training [24]. The mechanics of locomotor training involve a harness by which the patient is supported to partially unload his/her body weight while walking on a treadmill. Critical elements of this approach include: upright orientation relative to gravity, weight-bearing through both limbs, and the opportunity to utilize a bilateral, reciprocal gait pattern that involves hip extension [16, 22, 25]. Of note, each of these elements is a biomechanical characteristic of normal walking [26].

One of the major proponents of locomotor training [22] demonstrated remarkable and functionally significant improvements in walking function in hemiparetic individuals who initiated training from either a nonambulatory or minimally ambulatory state during the subacute period poststroke. These effects were observed in a series of single subject ABA designs where the A phases involved locomotor training and the B phase traditional, Bobath, gait therapy [27]. Improved walking ability and speed were observed in both locomotor training blocks but demonstrated a plateau in the traditional therapy block. These results illustrate the efficacy of the locomotor training approach, even in hemiparetic persons demonstrating extremely low levels of physiologic function. Further, they stand in contrast to those suggesting the need for minimal functional capacity to benefit from task-specific gait rehabilitation [21, 28].

15.4 Recipe for Success

15.4.1 Neuromechanical Constituents

Walking for successful community ambulation is a complex behavior requiring control to: (1) produce a bilaterally reciprocal stepping pattern with sufficient propulsion for steady-state ambulation, (2) maintain balance during forward propulsion/progression [29], and (3) adapt walking to the behavioral goals of the person and the constraints imposed by the environment [30]. All three of these subtasks of walking are compromised as effects of stroke. Locomotor training has been proposed as an effective approach to promote walking recovery because it offers the requisite task specificity to address these subtasks.

Locomotor training is based on a model for the neural control of walking and its functional requirements as described by Forssberg and adapted by Barbeau [31–33]. Locomotor training in neurorehabilitation originally emerged for persons with spinal cord injury (SCI). Motivated by relative similarities between animal and human models of SCI, much attention focused on the role of afferent information in generation of appropriate patterned muscle activity during stepping. Afferent signals from muscle spindles and load receptors are important for promoting a proper locomotor rhythm in the central nervous system [34, 35]. Animal models demonstrate that sensory inputs are both phase and task specific [36–38] and that loading and unloading cycles are important for activation of extensors during stance [39]. In human bipedal locomotion, the leg flexor and extensor muscles are differentially controlled with more centrally determined control of flexors and more peripheral afferent input influence on extensors [40]. The spinal locomotor pools are highly responsive to phasic segmental sensory inputs associated with walking and demonstrate evidence of learning during step training [41–44]. Walking speeds and postural challenges have proven essential to improve locomotor outcomes [31, 45], emphasizing the need for locomotor training to establish conditions that present and optimize the relevant and appropriate signals to the spinal locomotor pattern generators in order to induce activity-dependent plasticity.

A critical perspective of the locomotor training approach is recognition of the inherent capacity for plasticity, even following central nervous system insult. Locomotor training shifts the goal of the rehabilitation intervention from enabling compensatory mechanisms to promoting recovery of neurologic function [31, 33, 42, 46, 47]. Because the locomotor training approach focuses on restoring locomotor capacity [44, 48–53], it depends on establishing conditions in which the damaged nervous system can experience normalized movement patterns.

The combined efforts of both animal and human studies emphasize the importance of normalizing stepping kinematics to promote activation and relearning of appropriate motor patterns in the spinal networks [16]. With increased limb loading, as provided through locomotor training, increased muscle activity (EMG) is observed in leg extensors during stance and, reciprocally, in contralateral leg flexors during swing [54, 55]. In the post-acute recovery period, continuous, nonspecific EMG patterns are typically observed during both standing and stepping [16, 56], but over time, by promoting appropriate stepping patterns through the physical assistance of partial body-weight support, treadmill, and manual advancement of paretic limbs, appropriately timed patterns of EMG bursting emerge. With training, these bursts become progressively refined and increasingly efficient, as they are specifically tuned to the gait phase. Sullivan and coworkers [56] observed both the emergence of phasic EMG patterns and significant reorganization of cortical activation [57] following locomotor training. These investigators observed reciprocal paretic limb dorsiflexion/plantarflexion in hemiparetic individuals was initially accompanied by weak cortical activation distributed diffusely across the leg and foot representation of sensorimotor cortex. Following locomotor training, central activation became more specific, as characterized by intense activation in a distribution focused on the contralateral foot region [57]. Related work using near infrared spectroscopy (NIRS) to monitor cortical activity during walking [58]. Miyai and coworkers [58] studied individuals poststroke and found that cortical activation during unconstrained treadmill walking was initially asymmetrical, favoring the contralesional hemisphere. With partial body-weight support, gait performance improved toward normal proportions of stance and swing phases and became more symmetrical between limbs. Concomitantly, cortical activation was globally reduced with improved hemispheric symmetry. Regionally specific effects included significantly reduced activation of sensorimotor cortex, somewhat reduced activity in supplementary motor cortex, and increased activity in premotor cortex. Thus, there is evidence to support that locomotor training affects not only the spinal locomotor pattern generators but provokes supraspinal reorganization specific to the differential control of flexor and extensor muscles that characterizes human locomotion.

15.4.2 Establishing the Appropriate Environment

The concepts that underlie locomotor training are relatively simple, but the pragmatics of its delivery are less straightforward. Several factors, which include: limb advancement during swing phase, effective loading, means to experience normalized movement patterns, and movement variability, interact with the fundamental components of body-weight support and treadmill speed to influence overall training efficacy. The majority of studies to date report adjustment of the two fundamental parameters, body-weight support and treadmill speed, empirically, without providing a clear rationale based on either the biomechanics of walking or physiological function [59]. Specific to persons poststroke, adjustments of body-weight support are noted to influence: single-limb support time – especially of the paretic limb, upright posture, maximal hip and knee extension angles, and plantigrade orientation of the foot–ankle complex at weight acceptance. Adjustments of treadmill speed are noted to influence: cadence, stride length, muscle activation, and heart rate, and in combination reveal greater metabolic efficiency of training at higher speeds (reviewed in [60]). Taken together, the available evidence suggests that optimal levels to promote interlimb symmetry range between 15% and 30% body-weight support. While specific recommendations for treadmill speed are less tangible, the objectives remain to improve the timing and magnitude of relevant muscle activation during walking and promote independent and efficient overground walking at physiologic speeds.

Limb advancement during swing phase is not specifically influenced by the basic locomotor training parameters. Various investigators have addressed this problem using: manual assistance of one or more therapists/trainers [22, 56, 61, 62], electrical stimulation to trigger a flexion withdrawal response [63], neuromuscular electrical stimulation [64], restraining the treadmill to speeds that allow the patient to advance the limb independently [65, 66], and providing external support via handrail hold [67]. Robotic devices contribute in this regard, not only by assisting limb advancement during swing phase, but by inducing a physiologic gait pattern with appropriate timing of stance and swing phases [68–70] and phasing of interlimb coordination.

Understanding the relative and concurrent effects of adjusting parameters of the locomotor training environment is critical to the development of effective training paradigms for restoration of locomotor function. Here, it is important to recall that walking is a bilateral, cyclic behavior. Approaches that focus on particular gait deficit (i.e., so-called foot drop) or specific phase of the gait cycle (i.e., swing phase) may neglect effects elsewhere in the gait cycle. For example, ankle dorsiflexion is often deficient, or even absent, during swing phase in persons poststroke. However, singular focus on addressing deficient ankle dorsiflexion may neglect stance phase deficits that compromise the ability to position the limb effectively and achieve hip extension in terminal stance and ankle plantar flexion at the stance-to-swing transition. While a singular focus on promoting dorsiflexion during swing may address "foot drop," it is equally likely to induce compensatory movement strategies including excessive hip flexion, shortened step length, and increased cadence. In contrast, promoting appropriate hip positioning permits appropriately timed ankle power at the stance-to-swing transition and sets in motion a series of events that include normalization of paretic limb swing time and enables limb shortening via combined adjustments at the hip, knee, and ankle [7, 71]. Robotic approaches are unique in their capacity to address the multifactorial problem of human walking.

While there is enthusiasm for the locomotor training approach and the collective evidence to date suggests it promotes improved walking function, it is well recognized that locomotor training is labor intensive. As discussed previously, an important objective is to normalize the kinematics during bilateral stepping in order to elicit appropriate activity in the spinal circuitry [16]. However, it is difficult to produce repeatable kinematics within individuals during training, making it difficult to produce consistent results across individuals. Inconsistencies in motor performance of the therapists/trainers assisting movement impede presentation of normalized kinematics and repetition of consistent patterns during training. Indeed, the results of 15 randomized controlled trials comparing locomotor training and traditional gait training (i.e., overground gait training, motor relearning) in persons post-stroke reveal conflicting evidence regarding the efficacy of locomotor training [3, 72]. Perhaps more importantly, these mixed results highlight the difficulty in interpreting the effectiveness of manually applied cues during repetitive stepping. Considered in combination with the costs of delivering locomotor training [73], these challenges contribute to its limited acceptance in the clinical setting to date. In response to these challenges, robotic devices, such as the Lokomat® (Hocoma, Inc., Zurich, Switzerland), have been developed in an effort to automate locomotor training and offer more cost effective and labor efficient locomotor rehabilitation [68].

15.5 Robotic Approaches

15.5.1 Design Considerations

Important to consideration of the role of robots in rehabilitation is the designed intent of the device. For example, one of the earliest reports of a robot designed for rehabilitation of walking was REHABOT [74]. This device addressed the challenge of early mobilization in patients with multiple traumatic or orthopedic injuries by providing secure postural support and reduced weight-bearing during ambulation. Robotic control of these two parameters alone permitted early partial weight-bearing and ambulatory training for several hours per day in the acute hospital setting, and facilitated early return to independent walking. While the REHABOT may have lacked sophistication in the actual walking pattern, it did facilitate graded weight-bearing and physical activity during a critical period when patients would otherwise remain minimally active.

Another early entry into the rehabilitation robotics was the electromechanical gait trainer developed by Hesse and coworkers who reported remarkable success at improving ambulatory capacity in low functioning, nonambulatory individuals poststroke [69, 75]. Likewise, the Lokomat, developed by Colombo and coworkers, targeted attainment of some level of ambulatory capacity for otherwise nonambulatory individuals with incomplete spinal cord injury (iSCI) [68]. Considered from this perspective (i.e., promoting ambulatory function in nonambulatory individuals), the outcomes of robotic locomotor training are remarkable and significant.

As discussed previously, a clear strength of robotic training approaches is the ability for simultaneous control of the multiple parameters of walking. The attendant challenge in robotic design is to identify important parameters involved in generating, controlling, and training the locomotor pattern and to prioritize these parameters appropriately. For example, the Lokomat was originally designed to mimic manual locomotor training for persons with iSCI and thus incorporated the key elements of walking: upright positioning relative to gravity, weight-bearing through both limbs, and a bilateral, reciprocal gait pattern incorporating hip extension. Given that the goal was to induce ambulatory capacity in nonambulatory individuals, the generic set of parameters incorporated into this design may have been sufficient.

The importance of mechanisms contributing to disordered locomotor control varies somewhat by pathology. For example, in nonambulatory individuals with iSCI, the first priority is generation of afferent signals that converge on the spinal locomotor circuitry to facilitate stepping. In contrast, persons poststroke retain some capacity for locomotion, including the ability to step. While, ostensibly, the spinal circuitry remains intact poststroke, descending motor drive to the spinal pools is compromised. Further, walking with an asymmetrical, hemiparetic gait pattern returns afferent signals to the spinal circuitry that are both diminished and anomalous. Dysregulated sensorimotor integration may thus be far more detrimental to locomotor recovery poststroke than the absence of sensory signals to activate the spinal circuitry as emphasized in iSCI. Reintegration of accurate afferent signaling and descending motor drive at the level of the bilateral spinal circuitry is likely an essential requirement for walking recovery poststroke. Thus, effective locomotor rehabilitation for persons post-stroke must explicitly establish the biomechanical conditions that normalize coordinated bilateral motor activity. Repeated expression of this coordinated bilateral pattern will ultimately induce activity-dependent neural plasticity. To meet these goals and inform robotic designs with greater sophistication, necessitates a depth of understanding of the neuromechanics of walking for each clinical condition.

Table 15.1 summarizes currently existing robotic devices developed for locomotor rehabilitation [68, 69, 76–84]. Of note are the various design approaches and the elements emphasized in these designs. Early approaches to rehabilitation robotics controlled few parameters and offered limited adjustability. As previously noted, the REHABOT was designed to enable upright posture and support partial weight-bearing. The AutoAmbulator incorporated these fundamental elements adding mechanically guided reciprocal stepping. However, it offered no ability to adjust walking speed, cadence, or step length. Motivated by the goal to address principles of task specificity, the electromechanical gait trainer and Lokomat built on these rudimentary design concepts and incorporated adjustability of multiple parameters. Numerous devices, including both stationary and wearable exoskeletons, and designs incorporating various theoretical approaches to restoring locomotor control, now represent the rapidly evolving field of rehabilitation robotics. As the technical aspects of robotic design become more tractable and the design expertise expands, it becomes more critical to understand the neural control of locomotion, which elements of locomotion should be controlled and when and how these controls should be adjusted. For example, the recently developed Anklebot focuses solely on restoration of ankle dorsiflexion during swing phase and uses backdrivability, meaning the participant experiences less resistance when

Table 15.1 Existing robotic devices for restoration of walking function

Device	Design	Number of studies	Theoretical approach	Authors	Pub Year
REHABOT	Automatic device suspends patient in standing, provides secure postural support, and prescribed weight-bearing. No forward propulsion	3/3	Accurate support of body weight. Simplify walking and allow early mobilization for individuals with multiple comorbidities and/or requiring use of orthoses or walking aids. Target individuals difficult to train using traditional gait approaches including parallel bars	Kawamura, Ide, Hayashi, Ono, and Honda [79]	1993
Lokomat	Exoskeleton, driven gait orthosis	47/17	Based on motor learning principles of task-specificity and repetition with less therapist effort to set paretic limbs; establish physiological pattern.	Colombo, Wirz, and Dietz [68]	2001
Electromechanical gait trainer	Movement of footplates simulates stance and swing with a 60/40 ratio. Ropes attached to the patient control vertical and lateral movements of the center of mass in a phase-dependent manner. Device enables nonambulatory subjects to practice gait-like movement with minimal assistance. Some adjustability now available in step length and training speed. Partial body-weight support provided as is support of hand rail in front of patient	22/20	Task-specific repetition with less therapist effort to set the paretic limb	Werner, von Frankenberg, Treig, Konrad, and Hesse [69]	2002
AutoAmbulator	Exoskeleton	None	Simple mechatronic device offers upright positioning and reciprocal gait pattern. Enables early mobilization and initiation of walking therapy	Information derived from HealthSouth resource materials [83]	2002
LoPeS	Powered exoskeleton, actuated degrees of freedom include pelvic translation in horizontal plane, hip ab/adduction, hip flexion/extension, knee flexion/extension	3/1	Impedence control, assist-as-needed	van Asseldonk, Veneman, Ekkelenkamp, Buurke, van der Helm, and van der Kooij [80]	2008
Haptic Walker	End-effector; programmable footplate concept	1/1	End-effector principle	Hussein, Schmidt, Volkmar, Werner, Helmich, Piorko, Kruger, and Hesse [81]	2008
Tibion PK100	Wearable exoskeleton robot	1/1	Intention-based assistance/resistance during stance phase only	Horst [82]	2009
Powered ankle exoskeleton	Pneumatically powered exoskeleton	10/0	"Pneumatic muscles" augment and effectively increase plantar flexor strength during walking	Ferris [84]	2009
ALEX	Active leg exoskeleton and force-field controller	1/2	Assist-as-needed paradigm. Undesirable gait motion is resisted. Assistance provided toward desired motion. Effective forces applied at ankle through actuators at hip and knee	Banala, Kim, Agrawal, and Scholz [77]	2009
Anklebot	3-Degree-of-freedom backdrivable, wearable robot, actuated in sagittal and frontal planes	1/1	Dorsiflexion assist	Khanna, Roy, Rodgers, Krebs, Macko, and Forrester [76]	2010
G-EO-Systems	End-effector; programmable footplate concept	1/1	End-effector principle	Hesse, Waldner, and Tomelleri [78]	2010

Current robotic devices including brief treatment of design approach and device intention. Devices listed by publication date. Citations found from pubmed using terms "robotics, rehabilitation, stroke, gait training". Number of studies reflects: total citations for a given device (numerator) and studies reporting applications with persons post-stroke (denominator).

performing the desired motion [76]. In contrast, the ALEX device focuses on endpoint control [77] of foot–ankle placement using force fields to actively resist undesired motion and offering assistance as needed toward desired motion at all three joints: hip, knee, and ankle. As the field of rehabilitation robotics has matured, devices themselves and the control strategies have grown more sophisticated to incorporate: multijoint control to reproduce full kinematic trajectories of normal walking, multiple degrees of freedom at single joints (Hocoma, Inc., Zurich, Switzerland), physiologic variability [85], and assistance as needed [77, 86]. However, many questions remain regarding the control strategies. Invoking the principle of Occam's razor, the relatively parsimonious approaches of the REHABOT or the first generation Lokomat addressed only global aspects of normal gait (i.e., upright posture, reciprocal stepping, appropriate proportions of stance: swing or loading:unloading). But how much is enough? Did these relatively simple approaches afford sufficient physiologic specificity? Alternatively, is it necessary to develop subject-specific templates of the locomotor pattern? [87] Is this degree of sophistication necessary for certain neurologic conditions?

15.5.2 Current Evidence

Table 15.2 presents a brief synthesis of studies that have investigated acute (i.e., immediate, single-session) effects of robotic-guided walking in nondisabled individuals and persons poststroke or spinal cord injury [76–78, 80, 81, 86, 88, 89]. While it is not surprising that the device itself affects the spatiotemporal and kinematic characteristics of walking, these observations are important for identifying specifically how a robotic device influences the gait pattern. Further, this information provides important context for interpreting the outcomes of subsequent intervention studies and serves to inform the ongoing process of device development. For example, Neckel et al. observed healthy individuals while walking in the Lokomat and found that hip and knee angles and swing time are reduced but hip extension is increased [88]. These observations quantify expected differences between robotic-assisted and unconstrained treadmill walking; that is, the normal walking pattern is affected by moving against the robotic device. They also confirm the ability of the Lokomat to emphasize hip extension in terminal stance, which is a key objective of the locomotor training paradigm. Subsequent studies performed by these investigators involved individuals poststroke and revealed few significant differences in either kinematics or kinetics between healthy, paretic and nonparetic limbs during robotic-assisted walking. These observations indicate the capacity for the Lokomat to induce similar biomechanical effects between nondisabled and hemiparetic individuals. In reviewing the studies in Table 15.2, a common theme across devices is that nondisabled individuals demonstrate reduced joint angle excursions and increased swing time. Both effects are consistent with walking slower against an increased load.

Summarized in Table 15.3 are 11 studies drawn from the current literature that compare robotic locomotor training to either conventional rehabilitation, including gait training, or manual locomotor training for persons poststroke [1, 3, 69, 78, 90–96]. Characteristic of the rehabilitation literature, the study designs, therapeutic prescription, participant characteristics, and outcome measures vary tremendously, making it difficult to identify either distinct differences between training approaches or key elements where training approaches induce differential effects. Nine studies [1, 3, 90–95, 97] involve experimental designs, one study [96] involved biomechanical analysis of a subset of participants from one of the experimental designs [1], and one is a single case report [78]. Three experimental studies [91, 92, 95] were conducted in the acute rehabilitation period (i.e., <60 days poststroke), four involved persons in the subacute period (i.e., 2–10 months postevent) [69, 90, 93, 94], and two studied chronic hemiparetic individuals (i.e., >2 years postevent) [1, 3].

Table 15.2 Neuromechanical effects of robotic guidance

Citation	Device studied and device function	Population	n	Prescription	Study design	Outcome measures	Results
Neckel, Wisman, and Hidler [88]	Lokomat	Healthy	1	No intervention	Comparison of lower limb kinematics between Lokomat and treadmill walking at matched speed	Maximum knee angle Maximum hip angle Minimum hip angle	Maximum hip and knee flexion significantly lower in Lokomat compared to treadmill Maximum hip extension significantly higher in Lokomat compared to treadmill Percent time spent in swing significantly lower in the Lokomat
Neckel, Blonien, Nichols, and Hidler [89]	Lokomat	Stroke Chronic	10	No intervention	Comparison between controls, unimpaired, and impaired legs poststroke	ROM ankle, knee, and hip Max vertical pelvic displacement from heel strike Time of minimum pelvic displacement Maximum vertical ground reaction force Maximum ankle dorsiflexion torque Knee extension torque at midpoint of initial swing Time of maximum hip extension torque Hip adduction torque at midswing	Kinematics: no significant difference between control limbs and unimpaired limb of stroke subjects; No significant difference between the impaired and control limb; 1 significant difference between impaired and unimpaired limb (ankle ROM was less in the impaired limb). Kinetics: no significant difference between control and unimpaired limb of stroke subjects; 3 significant differences between impaired and control limb (maximum ankle dorsiflexion, knee extension at initial swing, and hip adduction at mid swing); 3 significant differences between impaired and unimpaired limb (maximum ankle dorsiflexion, knee extension at initial swing, and hip adduction at midswing). Lokomat torques: no significant difference between induced Lokomat torques on the three limbs (control, unimpaired, impaired)
Hussein, Schmidt, Volkmar, Werner, Helmich, Piorko, Kruger, Hesse [81]	Haptic Walker	Healthy	9	1	Comparison of free walking and stair climbing to each of 2 training modes; Comparison between 2 training modes	Mean normalized muscle activations during floor walking and walking up stairs	*Floor*: decreased activation shank extensor and flexor muscles, prolonged activation of thigh muscles, delayed onset from biceps femoris; *Stairs*: decreased activation of the shank muscles, (to a lesser degree), decreased activation of the thigh muscles and erector spinae, stronger activation of major weight-bearing muscles (when compared to floor walking)

Citation	device function	Population	n	Prescription	Study design	Outcome measures	Results
Emken, Harkema, Beres-Jones, Ferreira, and Reinkensmeyer [86]	Ambulation-Assisting Robotic Tool for Human Rehabilitation (ARTHuR)	Spinal cord injury (SCI) American Spinal Injury Association (ASIA) B–D Chronic	6	7 experiments within 2-h session	Proof of concept	Electromyographic (EMG) profiles Step height (kinematic data) Step length (kinematic data) Tracking error (kinematic data)	Kinematic trajectories: mean position tracking errors were within one standard deviation of the desired kinematic trajectories during walking
van Asseldonk, Veneman, Ekkelenkamp, Buurke, van der Helm, van der Kooij [80]	LOPES	Healthy	10	1	Randomized block design Types of walking: free walking on treadmill vs. walking with LOPES Walking velocities: 0.5 m/s 0.75 m/s 1.25 m/s	Stride time Total stance time Swing time Step length Single stance time Double stance ratio Step width Knee ROM Sagittal thigh movements Frontal thigh movements Frontal trunk rotation	All comparisons represent walking with LOPES vs. free walking Stride time: no significant difference Total stance time: no significant difference Swing time: significant increase walking with LOPES Step length: no significant difference Single stance time: Double stance ratio: significant decrease walking with LOPES Step width: significant increase walking with LOPES Knee ROM: reduced with LOPES walking Sagittal thigh movements: smaller with LOPES walking Frontal thigh movements: reduced with LOPES walking Frontal trunk rotation: significantly increased with LOPES walking
Banala, Kim, Agrawal, and Scholz [77]	ALEX	Stroke Chronic	2	Three, 5-day training sessions	Proof of concept	Tolerable treadmill speed. Ability of participant to track trajectory template developed from speed-matched healthy control. Template size	Tolerable treadmill speed: both participants showed improvements reflected by increases in their tolerable treadmill walking speeds Template tracking ability: both participants showed significant improvements in ability to track healthy control template Template size: both participants showed considerable improvements in template size and approximated control values

(continued)

Table 15.2 (continued)

Citation	Device studied and device function	Population	n	Prescription	Study design	Outcome measures	Results
Khanna, Roy, Rodgers, Krebs, Macko, and Forrester [76]	Anklebot	Stroke Chronic	10	No intervention[a] Anklebot was unpowered	Treadmill vs. overground and with vs. without anklebot on paretic leg	Kinematics Step time (footswitch data) Percent stance (footswitch data)	Spatiotemporals: no significant differences between overground and overground with robot; No significant differences between treadmill and treadmill with robot; improved symmetry with treadmill vs. overground. Kinematics: significant decrease in maximum paretic dorsiflexion during overground with robot vs. overground; nonparetic knee flexion greater in treadmill vs. treadmill with robot condition; Maximum paretic hip flexion higher with treadmill than overground; Maximum paretic hip flexion higher with treadmill and treadmill with robot than overground with robot condition; Maximum nonparetic hip flexion greater in the treadmill vs. overground with robot condition. No other significant differences
Hesse, Waldner, and Tomelleri [78]	G-EO-systems Device based on end-effector principle	Stroke Ambulatory Subacute ≥20 m, ≥0.25 m/s, ≥10 stairs reciprocally; AD and handrails allowed 6–14 weeks	6	No intervention	Real vs. simulated floor walking and stair climbing	EMG	*Simulated floor walking*: delayed onset and prolonged activation of the vastus medialis and vastus lateralis, relative to real walking condition; gastrocnemius showed a phasic pattern in simulated walking vs. a tonic pattern in real floor walking; *Stair climbing*: activation patterns and amplitudes comparable across conditions for the thigh muscles, shank muscles demonstrated timely activation in the simulated condition for 3 participants, and the gastrocnemius pattern became more phasic

[a]Goal of this study was observation of potential effects due to mass of the device

Table 15.3 Effects of robotic devices on walking function and recovery post-stroke.

Citation	Device studied and device function	Population	*n*	Prescription	Study design	Outcome measures	Results
Werner, von Frankenberg, Treig, Konrad, and Hesse [69]	Electromechanical gait trainer enabled nonambulatory subjects to practice gait-like movement with minimal assistance; movement of 2 footplates simulated stance and swing inducing timing of a physiological gait pattern; ropes attached to the patient controlled vertical and lateral movements of	Stroke Nonambulatory Subacute	30	30 sessions 5× per week 6 weeks Net walking time: 15–20 min in addition to other therapies	Randomized controlled study with crossover design A-B-A vs. B-A-B A = 2 weeks of gait trainer therapy (0–2.5 km/h) B = 2 weeks of treadmill therapy with BWS (0–5 km/h)	Functional ambulation categories (FAC) Rivermead motor assessment(RMA) score (gross functions and leg and trunk section), Modified Ashworth (mAshworth) (ankle DF) Gait velocity: (10-m overground, maximum speed)	FAC: Group A improved more than group B; RMA: Both groups improved over time, no group differences; mAshworth : No group differences; Gait velocity: Both groups improved over time, no group differences
Tong, Ng, and Li [90]	Electromechanical gait trainer	Stroke Subacute	46	20 sessions 5× per week 4 weeks 20 min of gait training as part of 1.5 h of interdisciplin-ary rehabilitation program	Randomized controlled trial Control: Conventional gait therapy (CGT) Experimental Groups: Electromechanical gait trainer + Conventional Gait Therapy (EGT) Electromechanical gait trainer with Functional Electrical Stimulation + Conventional Gait Therapy (EGT-FES)	5M Walking speed Elderly Mobility Scale (EMS) Berg Balance Scale (BBS) FAC MI (leg) Functional Independence Measure (FIM) BI	5M: Significantly faster following EGT-FES than CGT at 2 weeks and following EGT and EGT-FES rather than CGT at 4 weeks EMS: Significantly higher scores following EGT and EGT-FES than CGT at 4 weeks BBS: No significant differences between control and experimental groups FAC: Significant difference between CGT vs. EGT and CGT vs. EGT-FES at 2 and 4 weeks; control group did not show comparable improvements to experimental groups MI: Significantly higher strength score from EGT-FES than CGT at 4 weeks FIM: No significant differences between control and experimental groups BI: No significant differences between control and experimental groups No significant differences between EGT and EGT-FES. However, effect sizes demonstrate differences in favor of EGT-FES

(continued)

Table 15.3 (continued)

Citation	Device studied and device function	Population	*n*	Prescription	Study design	Outcome measures	Results
Pohl, Werner, Holzgraefe, Kroczek, Mehrholz, Wingendorf, Holig, Koch, and Hesse [91]	Electromechanical gait trainer	Stroke	155	20 sessions	Randomized controlled trial	FAC	FAC: Significantly greater number of *group A* could walk independently at tx end and 6 month f/u;
		Nonambulatory		5× per week	Group A: 20 min Electromechanical gait trainer + 25 min conventional physical therapy (PT)	Barthel Index (BI)	BI: Significantly more people in *group A* attained ≥75 at tx end. Difference not maintained at follow-up.
		Subacute (<60 days)		4 weeks	Group B: 45 min conventional PT	Gait velocity: (10-m overground, maximum speed)walking endurance: 6 min walk test	*Group A* performed significantly better on walking velocity, endurance, mobility, leg power at tx end, but not at f/u
				45 min		Mobility: Rivermead Mobility Index Leg power: Motricity Index (MI)	
Husemann, Muller, Krewer, Heller, and Koenig [92]	Lokomat	Stroke	30	20 sessions	Randomized controlled pilot study	FAC	FAC: Both groups improved over time, no group differences;
		Subacute (≥28, ≤200 days)		5× per week	Control: 60 min conventional PT	Gait velocity (10-m overground, maximum speed)	Gait speed and cadence: increased while stride duration: decreased over time, no group differences;
				4 weeks	Experimental: 30 min robotic training + 30 min conventional PT	Spatiotemporals (cadence, stride duration, stance duration, single support time for both legs)	Paretic single support time: Experimental group showed significant increase over control group;
				60 min		Body composition Muscle tone (mAshworth)	BI and MI: Increased over time, no group differences; mAshworth: No change, no group differences;
						Leg power (MI) BI	Body composition: Control group increased body weight and fat mass, experimental group did not change body weight, but exchanged fat mass for lean body mass

Mayr, Kofler, Quirbach, Matzak, Frohlich, and Saltuari [93]	Lokomat	Stroke	16	45 sessions	Prospective, randomized, blinded, parallel-group trial	Modified EU-Walking Scale	Group 1 (ABA): Improved EU-Walking Scale, RMA, MRC, 6MTWD, following Lokomat (phases I and III), MI improved after phase I, Ashworth improved after phase III; no significant improvements noted during phase II
		0.5–10 months		5× per week	ABA vs. BAB	RMA (gross function),	Group 2 (BAB): Improved RMA, 6MTWD, MI following phase I, improved EU-Walking Scale, MRC, 6MTWD, and Ashworth following phase II, and 10 m walk time following phase III; walking speed did not significantly improve from baseline to end of treatment
				9 weeks	A = 3 weeks Lokomat	Gait velocity: 10-m overground	No significant differences noted between groups at baseline; therefore, differences result from differential training modes.
				Up to 30 min	B = 3 weeks conventional PT	6MTWD Muscle strength (Medical Research Council Scale (MRC) and MI) Muscle Tone (Ashworth)	Training with the Lokomat produced improvements in EU-Walking Scale, RMA, MRC, 6MTWD, MI, and Ashworth score; whereas training with conventional PT produced improvements in only RMA, 6MTWD, MI, and 10 m walk time
Hornby, Campbell, Kahn, Demott, Moore, and Roth [1]	Lokomat	Stroke	48	12 sessions	Randomized controlled study	Gait velocity (SSWS and FAST measured with Gait Rite)	*By Group*:
		Ambulatory		3× per week	Control: Therapist-assisted locomotor training		Control group revealed larger gains than experimental group for SSWS, FAST, and single-limb stance%-FAST. Differences were not maintained at 6 month f/u
		Chronicity		4 weeks	Experimental: Lokomat	Single-limb stance% (Gait Rite)	No group differences detected for single-limb stance%-SSWS, step length asymmetry-SSWS, and step length asymmetry-FAST.
		Robotic-assisted: 50 (+/-) 51 months		30 min		Step length asymmetry (Gait Rite)	*By Severity*:
		Therapist-assisted:73(+/-) 87) months				6MTWD – overground distance covered walking at SSWS for 6 min	Participants with moderate deficits made greater improvements than participants with severe deficits in SSWS following treatment.
		Baseline SSWS				Modified Emory Functional Ambulation Profile (mEFAP)	Participants with severe deficits demonstrated greater improvements in step length asymmetry-SSWS. No differences detected for single-limb stance%.
		Robotic-assisted: 0.45–0.19 m/s				BBS Frenchay Activities Index	
		Therapist-assisted: 0.43–0.22 m/s				SF36 – physical component	

(continued)

Table 15.3 (continued)

Citation	Device studied and device function	Population	*n*	Prescription	Study design	Outcome measures	Results
Hidler, Nichols, Pelliccio, Brady, Campbell, Kahn, and Hornby [94]	Lokomat	Stroke:	63	24 sessions	Randomized clinical trial	Self-selected walking speed (SSWS) – 5-M overground	SSWS: control group improved more than experimental group;
		0.1–0.6 m/s <6 months		3× per week	Control: Conventional therapy	Cadence (Gait Rite)	
		<6 months		8–10 weeks	Experimental: Lokomat	6MTW	6MTW: control group improved more than experimental group;
				45 min		BBS FAC National Institute of Health Stroke Scale (NIHSS) Motor Assessment Scale (MAS) RMA Frenchay Activities Index SF-36	No group differences detected for FAC, RMA, BBS, MAS, cadence
Lewek, Cruz, Moore, Roth, Dhaher, and Hornby [96]	Lokomat	Stroke	26	12 sessions	Randomized clinical trial	Hip and knee average coefficient of correspondence (HK-ACC)	No group differences detected for SSWS, cadence, stride length, kinematics, and extent of limb circumduction.
		Ambulatory; able to walk ≥10 m without physical assistance; self-selected gait speed <0.8 m/s		3× per week	Control: Therapist-assisted locomotor training	SSWS	No group differences noted for HK-ACC in either paretic or nonparetic legs.
		Chronic (>6 months) (subset from Hornby et al. 2008)		4 weeks 30 min	Experimental: Lokomat	Cadence Stride length Kinematics Extent of limb circumduction	Within-group improvement noted for HK-ACC paretic in the control group

Westlake and Patten [3]	Lokomat	Stroke	16	12 sessions	Parallel, randomized design (pilot study)	Gait velocity (SWSS and FAST measured with Gait Rite)	By Group: No significant differences between experimental and control group for SSWS, absolute paretic step length ratio, Fugl-Meyer, SPPB, BBS, LLFDI.
		Ambulatory		3× per week		Step length asymmetry	Experimental group: significant *within-group* improvements for SSWS, FAST, absolute step length ratio, Fugl-Meyer, SPPB, BBS; Control group: significant *within-group* improvements detected for only BBS.
		Chronicity		4 weeks	Control: Therapist-assisted locomotor training	6MTWD	*By Training Speed*: No significant differences noted between fast- and slow-trained groups on primary or secondary measures
		Lokomat: 43.8–26.8 months		30 min	Experimental: Lokomat	Fugl-Meyer	
		Therapist-assisted: 36.8–20.3 months				Short Physical Performance Battery (SPPB)	
		Baseline SSWS				BBS	
		Lokomat: 0.62–0.31 m/s Therapist-assisted: 0.62–0.28 m/s				Late Life Function and Disability Instrument (LLFDI)	

(continued)

Table 15.3 (continued)

Citation	Device studied and device function	Population	*n*	Prescription	Study design	Outcome measures	Results
Schwartz, Sajin, Fisher, Neeb, Shochina,	Lokomat	Stroke	67	30 sessions	Nonblinded prospective, randomized, controlled study	Ability to walk independently according to FAC scale	FAC: Experimental group showed significant improvement in ambulation ability (achieving FAC score ≥3). Control group did not.
Katz-Leurer, and Meiner [95]		Severity: 6–20 (NIHSS)		5× per week	Control Group: Conventional physical therapy	NIHSS	NIHSS: Both groups improved, greater improvement revealed in the experimental group.
		<3 months		6 weeks	Experimental Group: Lokomat	FIM	FIMcognitive: Both groups improved, no group differences detected. FIMmotor: Experimental group improved greater than control group.
				~48 min		Stroke activity scale (SAS)	SAS: Both groups improved, no group differences.
						Gait velocity – 10-m overground, maximum speed Timed Up and Go (TUG) Exercise tolerance – 2 min walk test Number of stairs climbed test	Tested for differences in gait velocity, TUG, exercise tolerance, and number of stairs climbed only in participants who achieved FAC ≥3. Only stair climbing revealed significant differences with the experimental group demonstrating greater improvement than the control group
Hesse, Waldner, and Tomelleri [78]	G-EO-systems	Stroke	1	25 sessions	Clinical case report	FAC	FAC: Improved from level 1 at baseline to level 4 at the end of 5 weeks; able to walk 20 m with a quad cane, without physical assistance
	Device based on end-effector principle	Nonambulatory		5× per week		MI	MI: Improved from a score of 22 at baseline to a score of 59 at the end of 5 weeks
				5 weeks		RMA	RMA: Improved from a score of 3 at baseline to a score of 7 at the end of 5 weeks
				25–30 min		BI	BI: Improved from a score of 25 at baseline to a score of 65 at the end of 5 weeks

Synthesis of studies investigating acute (n = 6) and intervention-related effects (n = 12) of robotic locomotor trainers/training.
Abbreviations: w/c - wheelchair, FAC – Functional ambulation categories, RMA – Rivermead motor assessment, mAshworth – modified Ashworth score, BI – Barthel Index, MI – Motricity Index, tx – treatment, f/u – follow-up, PT – physical therapy, CGT – Conventional Gait Therapy, EGT – Electromechanical Gait Trainer, EGT-FES – Electromechanical Gait Trainer with Functional Electrical Stimulation, EMS - Elderly Mobility Scale, BBS – Berg Balance Scale, FIM – Functional Independent Measure, MRC – Medical Research Council Scale, SSWS – self-selected walking speed, NIHSS – National Institute of Health Stroke Scale, MAS – Motor Assessment Scale, mEFAP – modified Emory Functional Ambulation Profile, SPPB – Short Physical Performance Battery, LLFDI – Late Life Function and Disability Instrument, HK-ACC – hip and knee average coefficient of correspondence, ARTHuR – Ambulation-Assisting Robotic Tool for Human Rehabilitation, SCI – spinal cord injury, ASIA – American Spinal Injury Association, EMG - Electromyography

15.5.3 Robotic Training Versus Conventional Therapy

Robotic training and conventional therapy were compared in six studies, all of which were conducted in the acute to subacute period of stroke recovery (i.e., range 28 days–10 months) [90–95]. The inclusion criteria for one study extended to the period from 6–10 months poststroke [93]. The robotic devices studied included the electromechanical gait trainer (EGT) [75] and the Lokomat [98]. Clinical measures of *impairment* [99] including the Rivermead Motor Assessment [100], Fugl-Meyer Test of Motor Function [101], NIH Stroke Scale, and either the Ashworth or modified Ashworth scale [102] reveal equivocal differences between training approaches. Two studies report no differences between groups [92, 94], while two studies favor robotic training [93, 95]. Indicators of *walking ability* including the: Functional Ambulation Categories (FAC) [103], EU-Walking Scale [104], Elderly Mobility Scale [105], and timed walking tests (6MTWD [106]) favor robotic training in four studies [90, 91, 93, 95], while two studies reveal no difference between robotic and conventional training [92, 94]. Three studies report improved *walking distance or endurance* following robotic training [90, 91, 93], while conventional therapy produced greater effects in one study [94]. Improvements on other indicators such as disability and activities of daily living (ADL) are difficult to assess and have not been consistently evaluated across studies. Nonetheless, two studies report gains following robotic training [91, 95], while two studies report no differences between approaches [90, 92]. Specific *gait parameters* also reveal mixed results. Two studies report greater improvements in walking speed following conventional therapy [93, 94], while Pohl [91] reported greater effects following training with the EGT and Husemann's investigation [92] revealed improvements following both approaches. Of note, following Lokomat training, Husemann [92] reported improved paretic single-limb support time during gait while Schwartz [95] reported improved ability for stair climbing. Both of these findings suggest improvements in strength, or power, particularly in the paretic limb. In this light, strength, broadly defined, using assessments including the Motricity Index [107], MRC Scale [108], or direct measurement of strength/power clearly favor robotic training [90–93]. In addition, Husemann's finding of increased lean body mass following robotic, but not conventional, training are noteworthy [92].

An important detail to note in the studies comparing robotic gait training to conventional therapy is that in many of these studies [90–92, 95], the experimental treatment involved both conventional gait therapy and robotic training. Thus, the comparison in these studies was not truly between robotic and conventional therapy. Rather, the study designs held the time in therapy constant and compared conventional therapy, including gait training, to a similar amount (i.e., time) of combined robotic and conventional therapy. Recognizing this critical detail, it is important to note that the actual amount of robotic therapy in these studies was half, or less than half, of the full time spent in each therapy session. Clarifying these parameters underscores the efficacy of robotic training and further confirms that the combination of robotic training and conventional physical therapy or gait training increases the likelihood of regaining independent ambulatory status [109]. These findings suggest that by combining robotic and conventional therapy, participants may be better able to consolidate and generalize locomotor adaptations, at least as probed by the clinical measurements used. Related to this point Mayr [93] used an alternating treatment design (i.e., ABA and BAB), which also combined robotic and conventional therapies, although presentation of the treatments was interleaved in blocks rather than within sessions. While Schwartz [95] compared robotic (Lokomat) to conventional gait training, the study design involved gait training sessions (3 per week) in addition to regular physical therapy daily for 6 weeks. Again, the significantly greater gains in neurologic status, ambulation capacity, and motor function revealed by the Lokomat-trained group were attained in conjunction with a regime of regular, conventional physiotherapy.

In a true parallel design, Hidler and coworkers [94] compared conventional therapy to robotic locomotor training with the Lokomat. While similar to studies discussed above, participants were in the subacute phase poststroke, but an important difference is that ability to walk without physical assistance was required for study inclusion. In contrast to studies discussed thus far, the conventional therapy group outpaced the Lokomat group on the primary outcome, overground walking speed. Secondary outcomes including walking ability, balance, and motor impairments revealed no differences between groups. These findings emphasize two important points regarding the efficacy of robotic locomotor training poststroke: first, robotic approaches demonstrate efficacy for improving ambulatory capacity; second, the strongest effects of robotic training have been demonstrated in participants with low levels of walking function (i.e., gait speeds <0.3 m/s [21]).

15.5.4 Robotic Training Versus Locomotor Training

A second set of studies involves comparisons between locomotor training, either with or without therapist assistance, and robotic locomotor training. Werner [69] utilized multiple ABA or BAB designs, where "A" phases involved the electromechanical gait trainer (EGT) in comparison to treadmill therapy with body-weight support ("B" phases) and revealed greater improvements in functional ambulation categories, but not walking speed, following training with the EGT in persons in the subacute phase poststroke [69]. In a study design with little contrast between experimental and control treatments, Hornby and coworkers [1] found that manual, or therapist-assisted, locomotor training revealed greater improvements in both self-selected and fast overground walking speed compared to an equivalent dose of locomotor training with the Lokomat. Of note, participants were in the chronic phase poststroke, and demonstration of ability to walk >10 m without physical assistance was required for study eligibility. Secondary outcomes revealing differences between treatment approaches include single-limb stance time in the fast walking condition and the physical function dimension of the SF-36. While statistical differences in single-limb stance were detected, it is important to note that the improvement reported was a 2% change, representing 20–22% of the gait cycle, while single-limb stance in healthy individuals is 39% of the gait cycle [110]. Taken together, these results suggest that both manual and robotic locomotor training improved walking speed, but did not appear to induce significant changes in the gait pattern. Instrumented gait analysis of a subgroup performed by Lewek and coworkers [96] revealed lack of change in kinematics, spatiotemporal parameters including cadence, step and stride length, and paretic limb circumduction in either group. Interjoint coordination, quantified using the hip–knee average coefficient of correspondence [111] (HK-ACC), revealed improved consistency following therapist-assisted locomotor training. While improved HK-ACC consistency was interpreted to reflect superior motor learning and skill acquisition and attributed to greater variability in the therapist-assisted condition, this interpretation warrants caution [112]. A higher HK-ACC indicates greater consistency in the interjoint coordination pattern, which may reflect strengthening of dysfunctional or aberrant locomotor coordination. Westlake [3] compared Lokomat and manually assisted locomotor training in chronic individuals poststroke. While the study design was quite similar to that utilized by Hornby, results of this pilot study revealed significant within-group differences favoring Lokomat training on primary outcomes of self-selected overground walking speed, fast walking speed, and step length symmetry. Potentially important differences between Westlake's and Hornby's studies include both chronicity poststroke and baseline walking function. Additional factors include specific training parameters. The Lokomat used by Westlake and coworkers was capable of attaining normal, physiologic walking speeds (i.e., 5 km/h or 1.4 m/s) [3]. Training in the range of speeds exceeding the standard Lokomat (i.e., 3.3–5 km/h) may have contributed to positive gait speed outcomes in the robot-trained group. Additionally, participants in this study were trained without ankle–foot orthoses that alter ankle joint range of motion and affect plantigrade orientation during limb loading, and when safely possible, Lokomat elastic foot lifters were removed to enable normal foot–ground interaction at loading and terminal stance. All means were observed to optimize position and load-related sensory signals that may influence the gating of spinal locomotor patterned activity.

Three of these studies tested persistence of treatment effects at times distal to the intervention. In all three cases, significant differential effects were detected immediately posttreatment [1, 91, 94], favoring robotic training [91], conventional therapy [94], and manual locomotor training [1] in one case each. Pohl's study, in subacute individuals, revealed differential treatment effects that were lost at follow-up 6 months postintervention [91]. It is important to note that treatment-related gains were retained, and even somewhat advanced, during follow-up, however, due to intersubject variability, statistical differences between groups were not detected at follow-up. Such results are quite typical of clinical research in rehabilitation. In contrast, Hornby's study involved individuals in the chronic phase poststroke (i.e., 4–6 years postevent). While revealing modest gains in gait speed and differential treatment effects immediately following locomotor training, these gains dissipated and differential treatment effects were no longer manifest at the 6 month follow-up evaluation [1]. Differences between conventional and Lokomat training were both greater and maintained statistical significance at 3 month follow-up in subacute individuals studied by Hilder [94]. Taken together, the available evidence demonstrates a robust biological effect of improved walking capacity that appears to be mediated by training and persists, to some degree. These effects are considerably greater when intervention occurs earlier in recovery. Robotic approaches appear to be particularly beneficial for promoting ambulatory ability in low level, or nonambulatory, individuals, especially in the acute phase of recovery. Once ambulatory capacity has been achieved, improvements in locomotor function, including changes in the locomotor pattern, have not been well investigated.

An additional detail to consider is that the majority of these studies involved greater, and more consistent, therapeutic doses relative to those offered in the typical clinical setting. Independent of chronicity poststroke, the doses tested in these studies reveal robust biological effects of improved walking speed and walking function generally consistent with fundamental principles of neural plasticity [113]. Given the fundamental rationale to offer more repetition and contextualized in the overall developmental timeline, these cumulative findings using rehabilitation robotics are encouraging. Devices have been developed. Their feasibility, safety, and fundamental efficacy have been demonstrated. Having attained these milestones, the challenge for the next generation of rehabilitation robotics is development of approaches that optimize therapeutic efficacy. Rather than mimicking current, conventional therapies, robotics holds potential to produce better, longer lasting effects more efficiently. But we need to understand the unique opportunities and parameters afforded by the robotic environment.

15.6 Can We Change the Fundamental Locomotor Pattern?

The capacity to restore the fundamental locomotor pattern in persons poststroke remains an unanswered question in neurorehabilitation. Once this capacity is revealed, there is a need to understand the most effective approach to locomotor restoration, and this information will inform the next generation of robotic designs.

15.6.1 Task-Specific? How Specific?

As discussed above, a critical perspective of the locomotor training approach is recognition of the inherent capacity for plasticity, even following central nervous system insult. Because locomotor training shifts the goal from attainment of walking capacity regardless of locomotor strategy, it is critical to establish conditions in which the damaged nervous system can experience normalized movement patterns. Locomotor training has been proposed as an effective approach to promote walking recovery because it offers the requisite task specificity to address the functional and biomechanical subtasks of walking.

Here, it is important to note that the current evidence remains inconclusive regarding whether locomotor training produces superior outcomes to traditional therapeutic approaches for persons poststroke [1, 94, 114, 115]. This lack of conclusive findings suggests the three putative walking functions: stepping, balance, and adaptability, as identified for animal and spinal cord injury models and described above, may not encompass all critical elements of the locomotor training paradigm as it relates to persons poststroke. Because both supraspinal and spinal segmental structures remain at least partially intact and patent, dysregulated sensorimotor integration may be far more critical than generation of sensory signals to activate the spinal circuitry, as is the goal in models of spinal cord injury. Our perspective thus holds that integration of afferent signaling and descending motor drive at the level of the bilateral spinal circuitry represents a fourth essential requirement for walking. Because the neural mechanisms controlling sensorimotor integration are disrupted poststroke, effective locomotor rehabilitation must explicitly establish physical and biomechanical conditions that normalize coordinated bilateral motor activity. Repeated expression of the coordinated bilateral pattern is necessary to induce activity-dependent neural plasticity. We assert that robotic approaches afford means to control the requisite physical and biomechanical parameters of walking and present normalized movement patterns to the damaged nervous system.

15.6.2 The Robot Is Not Passive

Contrary to prevailing expectation, robotic-guided locomotion is not passive. While EMG patterns differ somewhat between unconstrained walking and walking in the Lokomat, these differences are understandable given constraints of both treadmill walking and the presence of the robotic exoskeleton [116]. Speed-related modulation of EMG patterns during Lokomat walking in healthy individuals up to and including physiological walking speed (i.e., 1.4 m/s) [116] is consistent with speed-related scaling of EMG patterns during unconstrained treadmill walking [117]. Importantly, patterns at slow walking speeds (<3.5 km/h or 0.9 m/s) illustrate prolonged, and often ill-timed, muscle activation. However, as physiologic walking speeds are attained (>3.5 km/h), EMG patterns become progressively tuned and appropriately timed to the specific biomechanical functions of gait. Additionally, distal muscles, both tibialis anterior and gastrocnemius, are activated appropriately; thus, lack of actuation at the ankle does not impair the normal control strategies or the foot–ground interaction.

Our early experiences with the Lokomat revealed EMG activity generally consistent with expected timing of muscle activation patterns fulfilling biomechanical functions of gait. Of note, however, is an incidental observation illustrated in Fig. 15.1. Our data illustrate that muscle activation at, or below, self-selected walking speed is markedly asymmetrical with substantially less activity in the paretic relative to the nonparetic limb. This finding was not surprising and consistent with our earlier research [118] that revealed profound, disruptive influences of the nonparetic limb on paretic limb activation during bilateral, reciprocal locomotor activity. However, with progressive increases in walking speed in the Lokomat, paretic limb activation systematically increased, while nonparetic limb activation systematically decreased. These observations illustrate three salient points. First, robotic-assisted locomotion is not a passive phenomenon. Second, EMG is actively modulated during a single session of robotic-assisted walking indicating that the locomotor pattern is influenced by adjustments in the biomechanical parameters, including walking speed. Third, the symmetry of EMG activity between the paretic and nonparetic limbs improves markedly with increased stepping speed. This increased activation may exert, at least partial, inhibition on the nonparetic motor

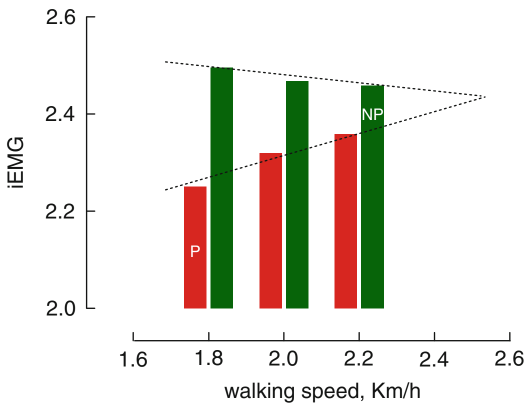


Fig. 15.1 Walking speed improves neuromotor symmetry. EMG data obtained from the vastus medialis of a chronic hemiparetic individual walking in the Lokomat at progressively increasing speeds. The Lokomat was operated using the default (bilateral position control) mode with 30% body-weight support. Foot lifters were used to assure limb/foot clearance. Treadmill speed was adjusted as tolerated. Quadriceps activity (integrated EMG per stride) becomes more symmetric with increased walking speed. Importantly, improved symmetry results from both increased paretic leg activity and reduced nonparetic leg activity and suggests a speed at which EMG will reach symmetry between limbs. These data clearly demonstrate that robotic-driven locomotion is an active, rather than passive, process

pools [119], producing a net normalizing effect on activation of the bilateral motor pools revealed as more symmetrical motor output. Of note, this phenomenon occurs simultaneously in multiple muscles at the same speed. Fundamentally, it is now possible to identify the subject-specific range of speeds where symmetrical neuromuscular activation is restored. Locomotor training in this range of speeds is likely to produce restorative effects on the locomotor pattern.

Another point related to activity during robotic walking was elegantly illustrated by Israel et al [120] in comparing metabolic cost (VO_2) between walking in the Lokomat and walking with manual assistance in persons with iSCI. Metabolic cost was markedly reduced while walking in the robot. This finding has been interpreted in favor of manual locomotor training, arguing that the intensity of exercise is greater during manual training and, further, that robotic-assisted walking is passive. However, considered in combination with changes in body composition reported by Husemann [92], these findings suggest that the lower metabolic cost of robotic-assisted walking may support sustained bouts of stepping with greater likelihood of inducing physiologic training effects. Moreover, metabolic cost was compared at matched speeds between manual and robotic-assisted walking. Manual locomotor training is typically conducted at the participant's "comfortable" walking speed. The ability to train for either sustained periods or at higher speeds, approaching normal walking speed, is limited by the capacity of therapists/trainers and, to some degree, discomfort of the participant. In this light, reduced metabolic cost during robotic-assisted locomotion offers: the potential to train at higher speeds, approaching physiologic walking speed; to experience more normal neuromotor patterns; and to sustain continuous stepping. It is noteworthy that Husemann found the control group (conventional therapy) increased body weight and fat mass over the 4 week (20 sessions) intervention, while the experimental (combined conventional and robotic training) group maintained body mass and exchanged fat mass for lean body mass. This difference may not be surprising, especially having made the point that studies investigating robotic training involve both increased dosage and consistency of dose, at least as defined by training time and, ostensibly, repetitions (steps), for all participants.

15.6.3 Altering the Biomechanical Environment

Spatiotemporal asymmetry between limbs is a hallmark of hemiparetic walking dysfunction. Illustrated in Fig. 15.2 [1, 3, 110] are differences in paretic single-limb support (expressed as percent gait cycle, SLS%) from 12 hemiparetic individuals during overground, treadmill, and Lokomat walking at the same speed. While it has been reported that treadmill walking, in and of itself, improves spatiotemporal symmetry [67], these data reveal that treadmill walking – without support of handrail hold – not only fails to improve SLS% symmetry but actually exacerbates asymmetry in some individuals. In contrast, walking with guidance of the Lokomat normalizes SLS% of both paretic and nonparetic limbs. Husemann's

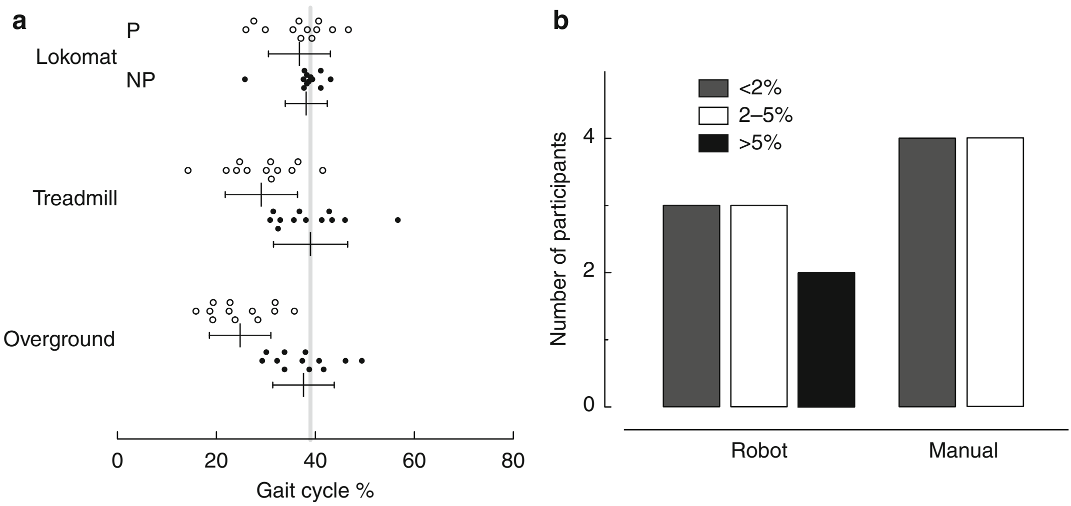


Fig. 15.2 (**a**) Single-limb support time across walking conditions. Single-limb support expressed as percent gait cycle for both paretic (*solid*) and nonparetic (*open circle*) legs in 12 chronic hemiparetic individuals during overground, treadmill, and Lokomat walking at matched speeds. Vertical cursor line at 39% of gait cycle denotes SLS% for normal, adult gait [110]. Individual subject data are presented with group mean (standard deviation) designated below each cluster. Asymmetry between paretic and nonparetic legs is obvious, and unchanged ($p>0.05$) between overground and treadmill conditions. During Lokomat walking, nonparetic limb SLS% is markedly normalized, clustering around 39% gait cycle. While variability among individuals remains present in the paretic leg, the group mean is markedly shifted toward normal ($p<0.01$), and means between limbs are similar ($p>0.05$). Note that 9/12 participants reveal P-SLS% near 39% gait cycle. (**b**) Changes in SLS% posttraining. Frequency counts of participants producing minimal (<2%), small (2–5%), or modest (>5%) improvements in P-SLS% (Data from Westlake and Patten [3], with eight participants per group (Lokomat and manually trained). Consistent with observations from Hornby et al. [1] changes following manual training are distributed between minimal and small magnitude improvements. The Lokomat-trained group demonstrated fewer individuals in the minimal change group and more improving P-SLS% >5%)

comparison of conventional gait training to combined Lokomat and conventional training revealed significant improvements in paretic single-limb support (P-SLS%) in the Lokomat-trained group [92] consistent with repetitive experience of loading the paretic limb for normal duration of SLS. While statistically significant differences between groups were not detected in the sample reported by Westlake [3], our data reveal more large improvements in P-SLS% (i.e., >5% of gait cycle) in participants who trained in the Lokomat (Fig. 15.2b). This finding is consistent with the biomechanical task specificity of normalized SLS% and loading experienced during Lokomat-guided training.

15.6.4 Are We Measuring the Right Outcomes?

Addressing the question of capacity for neuromotor recovery assumes we are measuring the appropriate outcomes. Few studies to date have probed beyond gross measures of walking speed or clinical outcome scales of gait ability, to determine whether, and how, locomotor training affects the neurobiomechanical walking pattern [22, 56, 65, 69, 112, 121]. The overwhelming majority of rehabilitation studies use overground walking speed as their primary outcome [3, 62]. While overground walking speed does reflect certain aspects of hemiparetic severity and functional capacity, its use as a primary outcome can be problematic because many factors contribute to walking speed. Improved walking speed can result from: physical conditioning, acquisition of compensatory movement strategies or genuine changes in locomotor function – either changes in coordination or neuromechanical function. Further, while many studies report small, perhaps clinically meaningful [122, 123], changes in overground gait speed, it is important to recognize the heterogeneity of response contributing to these group effects. Any of the existing literature reporting gait speed changes is likely reporting combined effects of responders, nonresponders, and even negative responders. Mixing these patterns of response obscures the ability to identify actual physiologic changes. To identify these differences requires measures with greater sensitivity.

15.6.5 Vertical Ground Reaction Forces

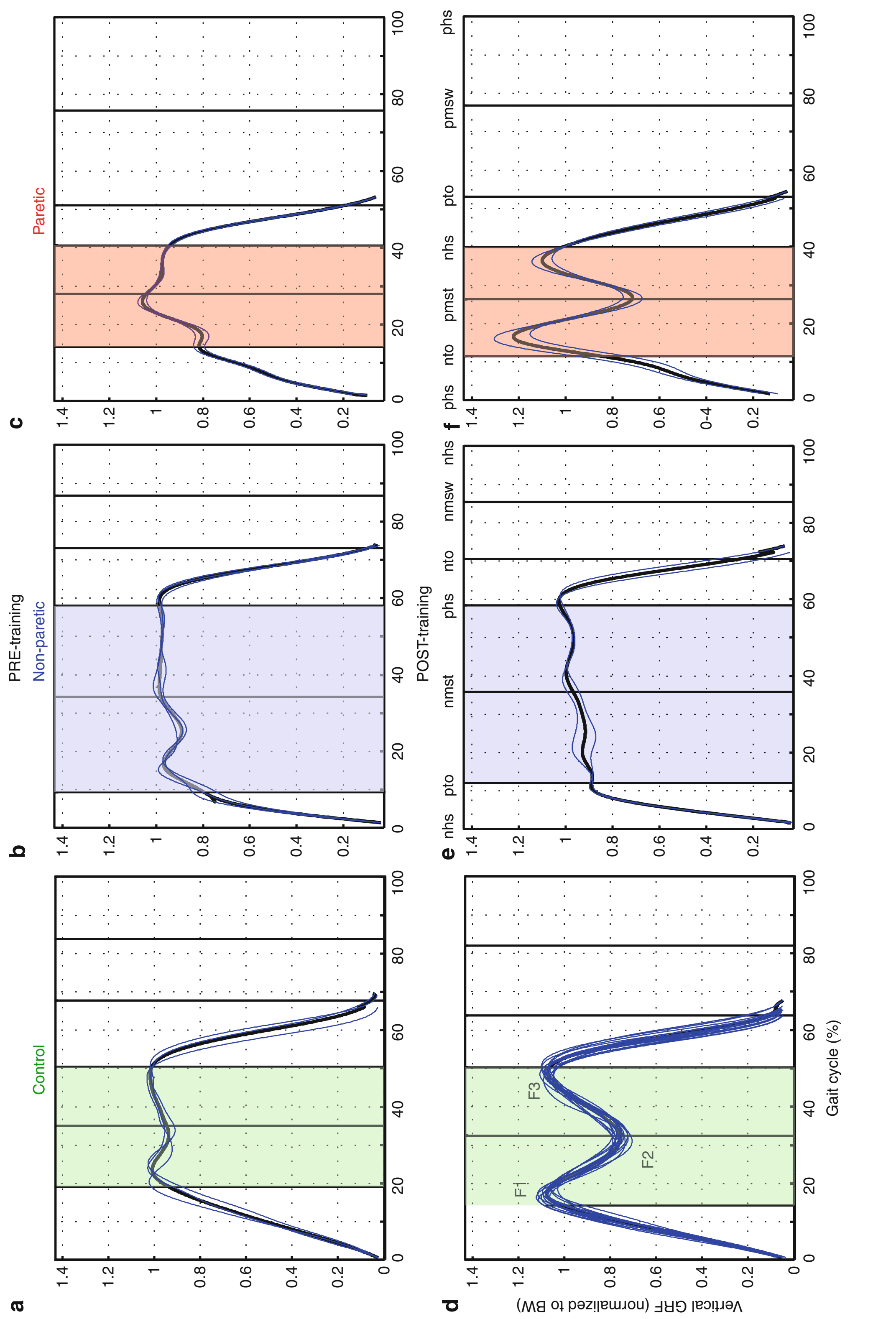
While it is important to understand effects induced while walking in the robot, it is critical to determine whether these produce persistent effects during unconstrained voluntary activity outside of the robot. Prior to and following our pilot study comparing robotic and manual locomotor training [3], we conducted instrumented gait analysis to characterize the gait pattern (pretraining) and identify changes (posttraining). Importantly, all participants were studied walking overground. Post-training studies were conducted within 1 week following completion of locomotor training. Both pre- and posttraining data are interpreted relative to reference normal by making comparisons between individuals with hemiparesis and nondisabled individuals walking at matched speeds.

Vertical ground reaction forces (Fig. 15.3) revealed some improvements consistent with not only increased and more symmetrical single-limb support, as described above, but improved loading and transfer of body weight between limbs. In addition to comparison between manual and robotic locomotor training, our study examined the effect of training speed. Half the sample was randomized to slow (<2.5 km/h or 0.69 m/s), while the other half was randomized to fast (>3.0 km/h or 0.83 m/s) training speeds. While our primary outcome (gait speed) did not reveal a significant effect, changes in the vertical ground reaction forces revealed improved symmetry between limbs, especially in individuals who trained either at fast speed or in the robot. For overall assessment of interlimb symmetry, we defined improvement as improved symmetry in at least 2 of the 3 peaks that characterize the vertical ground reaction force (i.e., F1, F2, and/or F3). Using this definition, the majority of participants who trained robotically demonstrated quantitative improvements in interlimb symmetry. Additionally, the majority of fast-trained participants demonstrated improved symmetry, while few such improvements were observed in slow-trained individuals. We also assessed intralimb changes in loading and unloading. Consistent with the interlimb effects discussed above, a greater number of improvements were observed in both paretic and nonparetic limb loading in the fast-trained individuals (Fig. 15.4a) [3]. Limb unloading patterns also improved somewhat in fast-trained individuals, although these effects were less dramatic (Fig. 15.4b). Although we anticipated the paretic limb would produce the greatest number of changes in vertical ground reaction force, our data revealed bilateral adaptations resulting from locomotor training. Training at fast speeds induced the greatest magnitude and number of improvements. Robotic configurations enable training at physiologic walking speeds for sustained periods and thus are critical to eliciting these effects.

15.6.6 Interjoint Coordination Patterns

We also investigated interjoint coordination patterns between the hip and ankle during overground walking to determine whether locomotor training alters the coordination pattern. Figure 15.5 illustrates our method for quantifying interjoint coordination (IJC). We compared IJC patterns in hemiparetic participants to control participants walking at similar speeds and identified positive changes when the hemiparetic subject's pattern became more similar to control between pretest and posttest. Likewise, we identified negative changes when subjects' patterns became more dissimilar to controls.

Tracking the centroid location of the hip–ankle angle–angle plot revealed that the majority of individuals demonstrated improved IJC in both nonparetic and paretic limbs. Interestingly however, nonparetic limb improvements appear to predominate from the hip. Further, our analysis detected differential patterns of improved IJC between Lokomat and manually trained individuals. Across all participants, we found that Lokomat-trained individuals improved IJC more


a
b
c
d
e
f
Control
PRE-training
Non-paretic
Paretic
POST-training
Vertical GRF (normalized to BW)
Gait cycle (%)
F1
F2
F3
nhs
pto
nmst
phs
nto
nmsw
pmst
pmsw

significantly than manually trained individuals. While our analysis detected roughly an equal number of beneficial and detrimental changes in IJC patterns among manually trained individuals, these changes were equally distributed across both the paretic and nonparetic limbs. We were surprised to find detrimental changes (i.e., worse IJC) bilaterally at the ankle suggesting a loss of normal coordinated ankle motion following manual locomotor training. Most notably, however, improved paretic limb IJC in Lokomat-trained individuals resulted from concurrent hip–ankle contributions. This pattern of concurrent joint contributions suggests that robotic training promoted reacquisition of the coordinated motor pattern rather than compensation for hemiplegic gait with exaggerated single joint contributions to IJC.

15.7 Conclusions – Ongoing Development and Future Thinking

Current perspectives in neurorehabilitation recognize the inherent capacity for neuroplasticity, even following central nervous system insult. Therefore, it is critical to establish conditions in which the damaged nervous system can experience normalized movement patterns, especially the sensory experience that stems from appropriate mechanical loading and movement. Robots could be used to our advantage in this regard, but to date, this has not been the overriding perspective. Our initial experiences with the Lokomat afford optimism that it is indeed possible to change (improve) the fundamental locomotor pattern in persons poststroke. More importantly, these findings belie capacity for neuromotor recovery that has otherwise gone unrecognized due to use of suboptimal outcome measures and has remained untapped due to inability to effectively induce appropriate neuromechanical conditions. As rehabilitation robotics move to the next generation of development, there are opportunities for continued technological advancements. Rather than replication of clinical effects, the goal and expectation of these future designs is neural recovery and restoration.

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Fig. 15.3 Representative vertical ground reaction forces (*vGRF*) during overground walking. Patterns in nondisabled individuals (**a** and **d**) illustrate a positive peak (F1) at ~12% of the gait cycle, representing the beginning of single-limb support (*SLS*), as the contralateral limb initiates swing, the negative peak (F2) occurs at midstance as the contralateral limb is in midswing; the second positive peak (F3), occurring at ~50% of the gait cycle, represents the end of SLS, as the contralateral limb begins stance. The shaded area in each plot represents SLS. The epoch between F1 and F2 represents limb loading as the center of gravity (*COG*) moves over the support limb. The epoch between F2 and F3 represents limb unloading as the COG translates forward of the support limb in preparation for swing. The magnitude of the F1, F2, and F3 peaks results from differences in walking speed, (**a**) (0.7 m/s) and (**d**) (1.16 m/s) correspond with walking speeds produced by a hemiparetic individual who participated in robotic locomotor training. (**b**) (nonparetic limb) and (**c**) (paretic limb) illustrate vGRFs at self-selected walking speed (*SSWS*, 0.66 m/s) prior to LT. Following LT in the Lokomat distinctive peaks in the paretic limb vGRF at SSWS are illustrated (**f**) indicating normalization relative to nondisabled individuals (Abbreviations: *nhs* nonparetic heel strike, *pto* paretic toe off, *nmst* nonparetic midstance, *phs* paretic heel strike, *nto* nonparetic toe off, *nmsw* nonparetic midswing, *pmst* paretic midstance, *pmsw* paretic midswing)

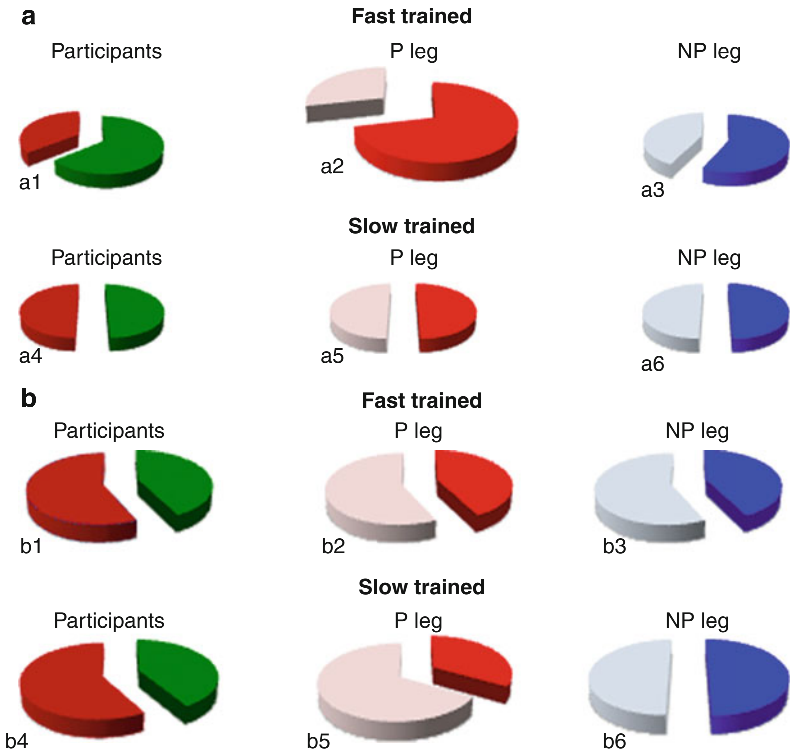


Fig. 15.4 (**a**) Limb loading (F2/F1). Changes in limb loading identified by analysis of vertical ground reaction forces (*vGRF*) obtained during overground walking following locomotor training in persons with chronic post-stroke hemiparesis (results reported in Westlake and Patten [3]). The ratio of the F2/F1 vGRF peaks characterizes loading and transfer of body weight onto the stance limb during the single-limb support phase of gait. Participants were stratified to slow (<2.5 km/h or 0.69 m/s) vs. fast (>3.0 km/h or 0.83 m/s) training speeds. Independent of manual or robotic training mode, the majority of fast-trained participants (64%) demonstrated improvements in limb loading (*a1*) which were noted more frequently in (*a2*) paretic (71%) vs. (*a3*) nonparetic (57%) legs. Fewer improvements in limb loading were observed in slow-trained individuals (50%) (*a4*) and were equally distributed across (*a5*) paretic and (*a6*) nonparetic legs. While improvements in limb loading were observed in fast-trained individuals following both manual and robot training modes, robotic training offers a clear advantage to achieve physiological walking speeds and maintain a coordinated stepping pattern. (Legend: *a1* and *a4 green* – improved vs. *red* – nonimproved; *a2*, *a3*, *a5*, and *a6* solid – improved vs. shaded – nonimproved) (**b**) Limb unloading (F2/F3). Changes in limb unloading identified by analysis of vertical ground reaction forces (*vGRF*) obtained during overground walking as described above in Fig. 15.3. (Data from participants as reported in Westlake and Patten [3]). The ratio of the F2/F3 vGRF peaks captures the single-limb support phase of gait from mid- to late stance and characterizes acceleration of the center of mass and transfer of body weight onto the contralateral limb. Improvements in limb unloading were revealed in both the (*b1*) fast (43%) and (*b4*) slow-trained (42%) individuals and were observed equally in (*b2*) paretic and (*b3*) nonparetic legs in fast-trained individuals. In slow-trained participants (*b5*), the paretic leg showed fewer improvements than the (*b6*) nonparetic leg. Across both fast and slow training speeds, the majority (82%) of improvements in limb unloading were revealed following robotic training. (Legend: *b1* and *b4 green* – improved vs. *red* – nonimproved; *b2*, *b3*, *b5*, and *b6* solid – improved vs. shaded – nonimproved)

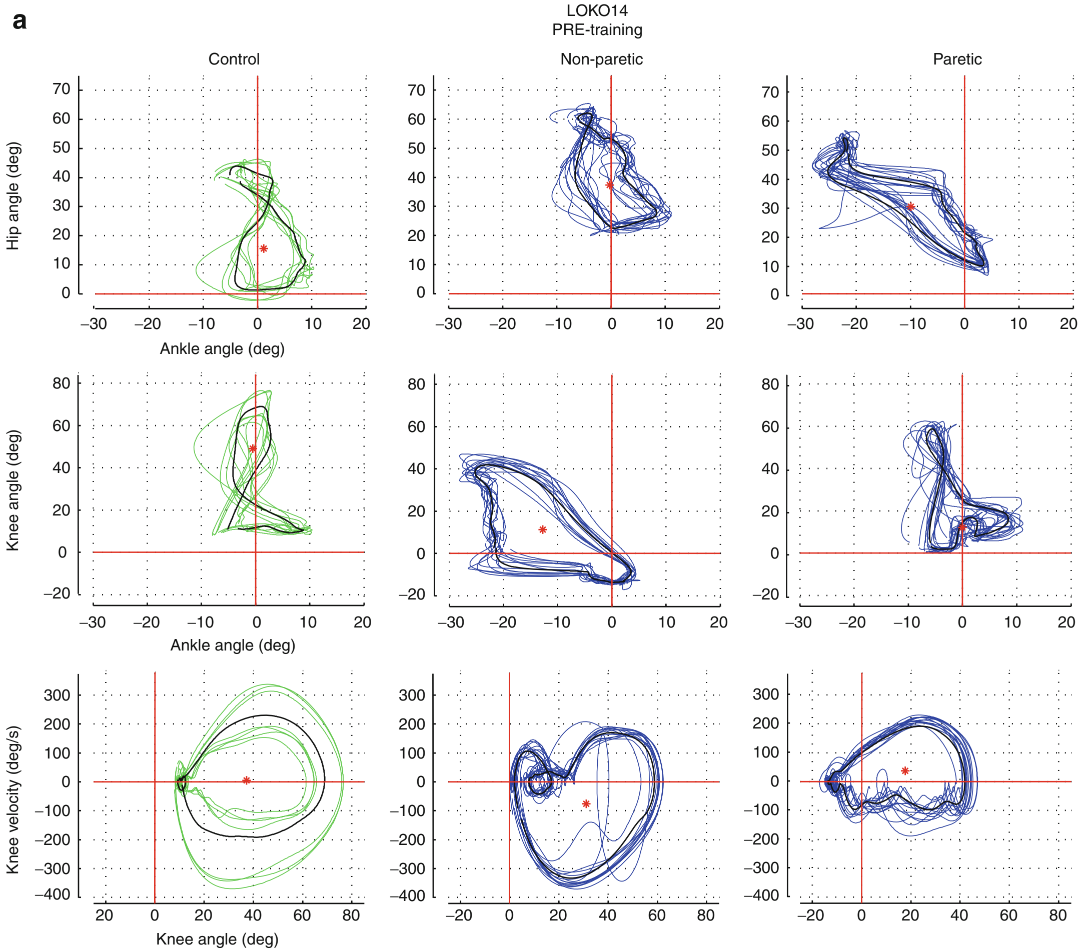


Fig. 15.5 Interjoint coordination. Coordination patterns derived from kinematics obtained during overground walking at self-selected walking speed. Top row: Hip–ankle angle–angle plots representing the excursions (*deg*) of the hip (*y-axis*) and ankle (*x-axis*) joints, respectively. Middle row: Knee–ankle angle–angle plots representing the excursions (*deg*) of the knee (*y-axis*) and ankle (*x-axis*) joints, respectively. *Bottom row*: Phase planes representing the angular velocity (*y-axis*, *deg/s*) vs. excursion (*x-axis*, *deg*) of the knee joint. Individual traces represent gait cycles and illustrate similarity of the coordination pattern over repeated cycles. *Left column*: Representative data from a control participant, walking at speed matched to hemiparetic participant, illustrated in green. *Middle and right columns*: Data from the nonparetic and paretic legs, respectively, of a hemiparetic participant. Calculation and interpretation of centroid location: The outer perimeter of the shape was used to calculate the centroid location (illustrated in red) and determine its coordinate location and distance from the origin. The absolute magnitude of the difference of centroid distance from origin is used to compare a participant to an individual, speed-matched control and to evaluate changes from pre- to posttraining. Movement of the centroid location toward control values is defined as a positive change. The centroid location can be decomposed into contributions from the x- and y-axes enabling identification of which joints (or joint) are deficient in their motion throughout the gait cycle and whether locomotor training induces changes in coordination.

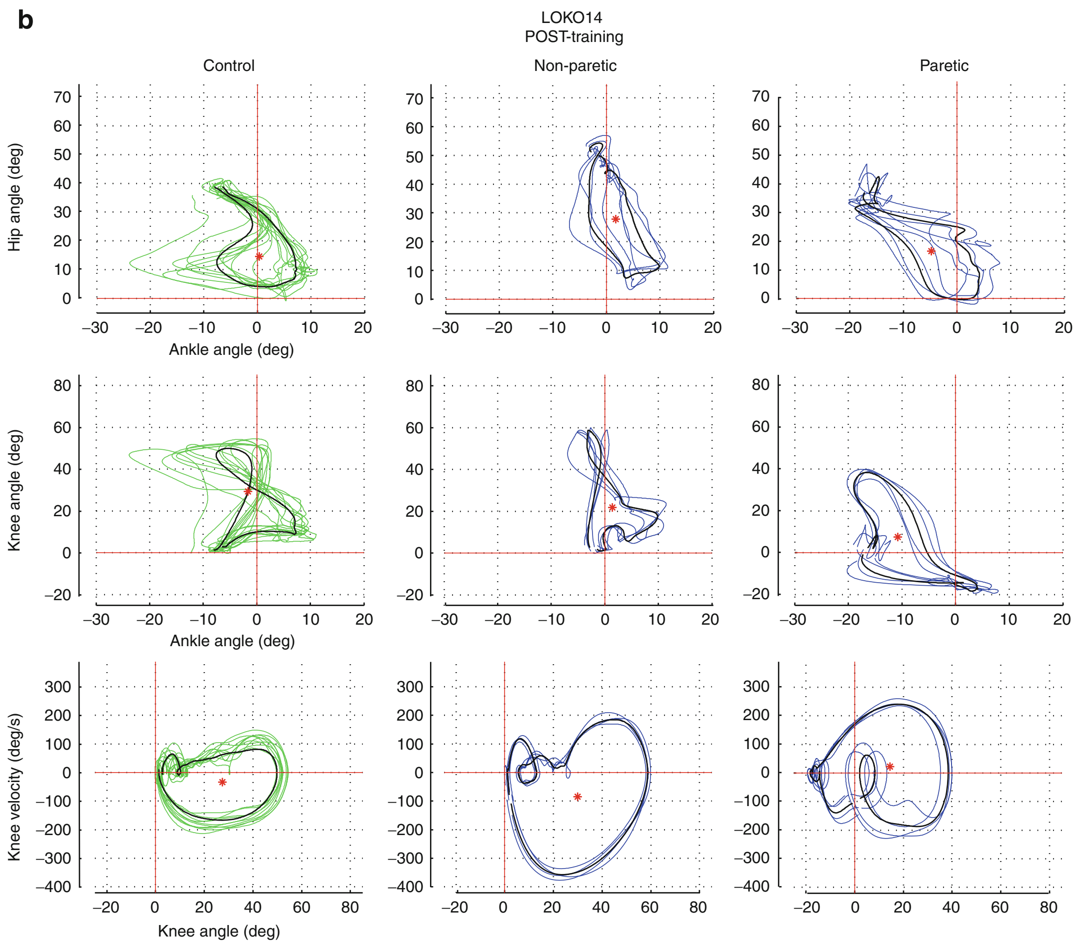
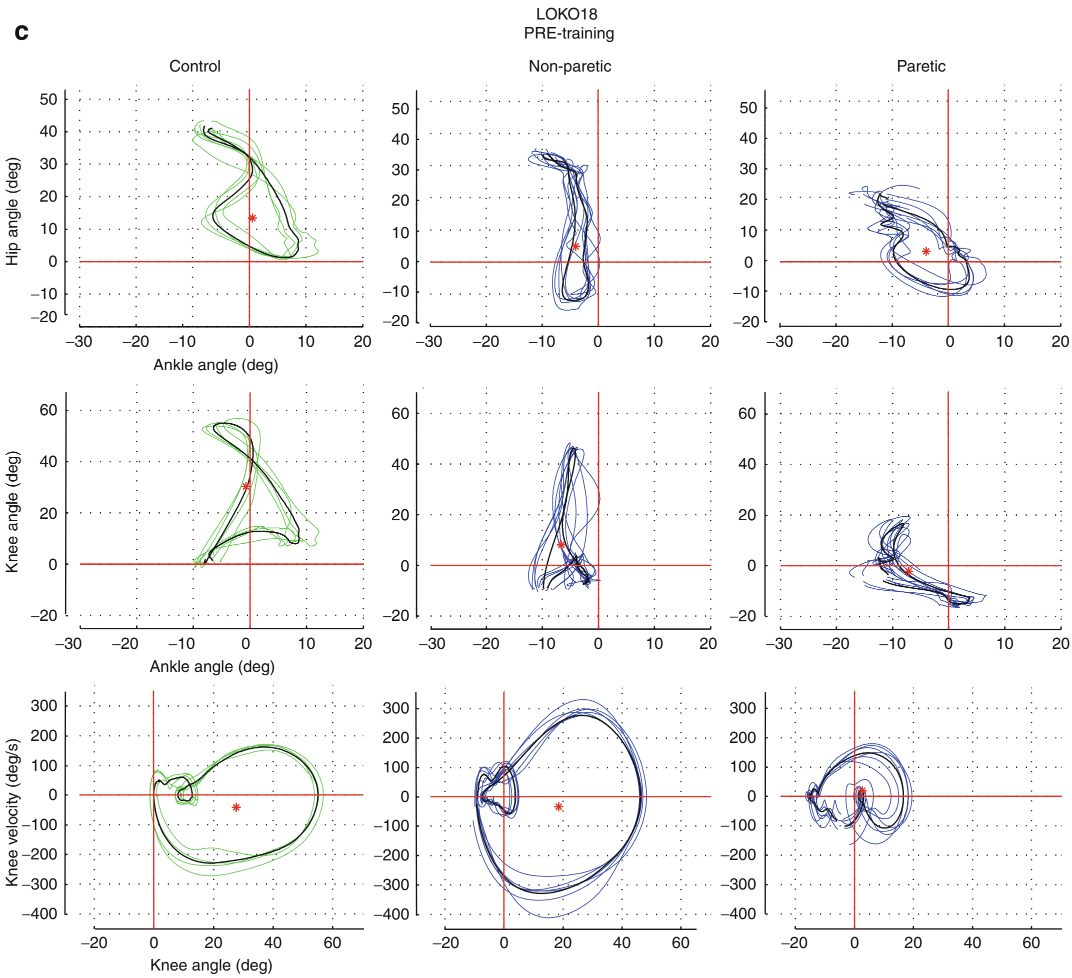


Fig. 15.5 (continued) (**a** and **b**) Interjoint coordination (*IJC*) patterns from a hemiparetic individual who trained with manual assistance. Self-selected walking speed (*SSWS*) = 0.44 m/s, absolute step length ratio (*SLRabs*) = 0.22. IJC patterns prior to LT (Fig. 15.5a) reveal bilateral deficiencies of knee and ankle excursion, compensated by exaggerated hip flexion, and compression of the knee joint phase plane. Posttraining, self-selected walking speed (*SSWS*) = 0.55 m/s, absolute step length ratio (*SLRabs*) = 0.14. Nonparetic limb IJC patterns reveal subtle improvements at the hip and marked improvements in knee–ankle coordination toward normal. However, paretic limb patterns reveal coordinative changes that suggest reduced excursion and poorer coordination across all joints. (**c** and **d**) Interjoint coordination (*IJC*) patterns from a hemiparetic individual who trained with the Lokomat. Self-selected walking speed (*SSWS*) = 0.69 m/s, absolute step length ratio (*SLRabs*) = 0.24. IJC patterns prior to LT (Fig. 15.5c) reveal minimal hip–ankle dyscoordination resulting from deficiencies of ankle excursion. Knee–ankle patterns are more aberrant with contributions from both joints. Posttraining (Fig. 15.5d), self-selected walking speed (*SSWS*) = 0.75 m/s, absolute step length ratio (*SLRabs*) = 0.17. Centroid shifts in the hip–ankle IJC pattern reveal contributions from both hip and ankle with ankle excursion in the range of normal. Changes in the knee–ankle IJC centroid location result primarily from improved ankle excursion



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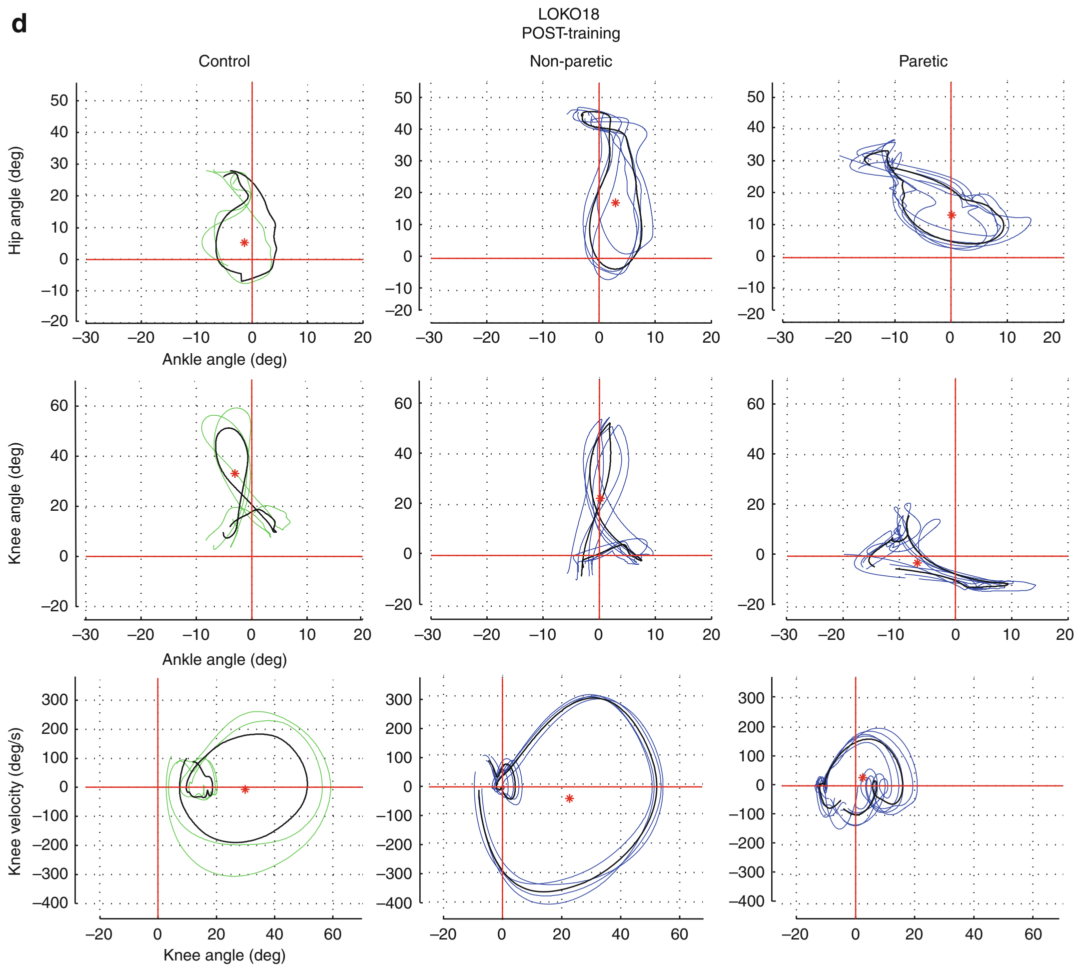

Fig. 15.5 (continued)

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