Guided by assistance, challenged by resistance

Minkes-Weiland, Sylvana

DOI:
10.33612/diss.1426327018

Publication date:
2025

Guided by assistance, challenged by resistance

Exploring effective strategies for gait rehabilitation after stroke

Sylvana Minkes-Weiland

The studies in this thesis were conducted within the Health in Context Research Institute of the University Medical Center Groningen (UMCG) and under auspices of the research program Smart Movements (SMART).

The printing of this thesis was financially supported by the Graduate School of Medical Sciences (GSMS) of the UMCG and the University of Groningen, Revalidatie Friesland and Stichting Beatrixoord Noord-Nederland.

Paranymfs: Martsen-Almer Minkes & Sara van der Velden
Cover illustration: Korja van der Werff
Design & printing: Proefschrift Maken

rijksuniversiteit
groningen

Guided by assistance, challenged by resistance

Exploring effective strategies for gait rehabilitation after stroke

Proefschrift

ter verkrijging van de graad van doctor aan de
Rijksuniversiteit Groningen
op gezag van de
rector magnificus prof. dr. ir. J.M.A. Scherpen
en volgens besluit van het College voor Promoties.
De openbare verdediging zal plaatsvinden op
woensdag 26 november 2025 om 11.00 uur

door

Sylvana Marijke Weiland
geboren op 21 oktober 1994

Promotores
Prof. dr. J.H.P. Houdijk
Prof. dr. L.H.V. van der Woude

Copromotores
Dr. A.R. den Otter
Dr. H.A. Reinders-Messelink

Beoordelingscommissie
Dr. J.M. Hijmans
Prof. dr. V.G.M. Weerdesteyn
Prof. dr. C.A.M. van Bennekom

Contents

CHAPTER 1

General introduction

Stroke impacts daily functioning

Stroke is characterized by an acute, focal neurological deficit resulting from vascular injury to the central nervous system. It expresses as either an infarction, accounting for approximately 85% of all strokes, or hemorrhage, responsible for the remaining 15% (Murphy & Werring, 2020). As one of the leading causes of disability worldwide, stroke places an increasing burden on healthcare systems. Between 1990 and 2019, the absolute number of stroke cases rose by 70% (GBD 2019 Stroke Collaborators, 2021). Currently, it is estimated that one in four people will experience a stroke during their lifetime (GBD 2016 Lifetime Risk of Stroke Collaborators, 2018).

In the Netherlands, stroke patients are first admitted to the hospital after the onset of symptoms, where initial acute rehabilitation begins (Zorgstandaard CVA/TIA, 2012). After initial hospital care, patients may receive either inpatient or outpatient rehabilitation in specialized hospital units or dedicated rehabilitation centers. For optimal recovery, the timely start of an intensive rehabilitation program is crucial, as a shorter time from stroke onset to rehabilitation is associated with better functional outcomes (Kwakkel et al., 1997; Maulden et al., 2005).

Stroke often has major functional and activity limitations that impact daily life. A common consequence is the development of unilateral motor deficits, most frequently in the form of hemiparesis (partial weakness) or hemiplegia (complete paralysis), which affect one side of the body, contralateral to the brain lesion. Functionally, motor impairments like hemiparesis and spasticity can restrict movement, while sensory deficits, cognitive issues, emotional and communicative challenges further affect basic functioning. At the activity level, these impairments make everyday tasks such as walking, self-care, and social interaction more challenging, potentially resulting in a loss of independence and reduced participation (Langhorne et al., 2011). Consequently, many stroke survivors experience a decline in their quality of life, struggling to return to work, engage in hobbies, and maintain meaningful social roles (Cramer et al., 2017). One of the most disabling symptoms after stroke are those involving the extremities which impact among others balance and functional gait (Nadeau et al., 1999). Problems in gait, that are initially present in more than 80% of the stroke survivors (Duncan et al., 2005), are often perceived as a major threat to ambulation, participation, health and quality of life (Schmid et al., 2007). Therefore, gait training has an important place in today's rehabilitation practice following stroke.

Problems in gait after stroke are multi-faceted

Upright human gait is a complex functional quality that is intensively learned early in life. It depends on motor functions and the ability to maintain balance. The gait pattern can be broadly divided into two primary phases: the stance phase (approximately 60% of the gait cycle) and the swing phase (approximately 40% of the gait cycle) (Perry & Burnfield, 2010). Each of these is further broken down into specific subphases. The stance phase comprises of loading response, mid-stance, terminal stance and pre-swing while the swing phase is subdivided into initial swing, mid-swing and terminal swing (Perry & Burnfield, 2010; Whittle, 2014). All these subphases involve a distinct aspect of the gait pattern that can be analyzed with measures of kinematics, kinetics and muscle activity. After a stroke, impairments such as muscle weakness, reduced motor control, and sensory deficits commonly disrupt these phases resulting in abnormal gait patterns. For example, patients often experience impaired shock absorption due to inadequate control of the quadriceps and plantar flexors leading to instability and inability to modulate forces during weight acceptance (Lamontagne et al., 2007). Toe clearance in swing is often reduced due to insufficient dorsiflexion, limited knee flexion (i.e. stiff knee gait) and hip flexion often resulting in compensatory movements like circumduction (Hsu et al., 2003). Moreover, due to affected calf muscles propulsion is often impaired (Turns et al., 2007). The distinct disruptions in the several gait phases often lead to an inefficient and asymmetrical gait that limits stability and mobility (Balaban & Tok, 2014), impacting the individual's independence and quality of life.

Although 64% of the people admitted for inpatient rehabilitation post stroke achieve independent walking before discharge (Jorgensen et al., 1995), yet gait abnormalities like a reduced gait speed, limited balance, a reduced knee flexion during swing, and in many cases a reduced push-off often persist (Lamontagne et al., 2007). Consequently, falling is a serious risk. Gait training strategies aim to restore as much function and independence as possible, enhancing overall well-being and participation (Shahid et al., 2023). However, the persistence of these gait deficits shows that current rehabilitation strategies are not yet sufficient for all patients, indicating a need for further improvement and individualization of gait training.

Gait training strategies post stroke

Substitution vs restitution

The gait pattern displayed by stroke survivors deviates from the normal pattern due to impaired function and compensation movements that depend on the individual's residual function (Balaban & Tok, 2014). As such, gait training programs can focus on either restitution or substitution of function. Restitution refers to the process of restoring impaired functions to their pre-stroke level through direct repair of damaged neural pathways or recovery of lost abilities (Kitago & Krakauer, 2013). Restitution is a key concept in for example Bobath/NDT therapy (Kollen et al., 2009) or constraint-induced movement therapy (Kwakkel et al., 2015). Substitution, on the other hand, involves compensating for lost functions by using alternative strategies, adaptive techniques, or assistive devices to achieve desired goals and tasks using an individual's remaining abilities (Kitago & Krakauer, 2013). Nowadays, rehabilitation programs focus more and more on substitution to achieve early functional independence (Kitago & Krakauer, 2013). However, a substitution approach during the early phases after stroke may limit long-term recovery. If spontaneous recovery is occurring, patients may unnecessarily rely on their acquired compensation behavior (Kitago & Krakauer, 2013). Therefore, the decision for a restitution or substitution approach should be based on information about the rehabilitation phase, the expectation for natural recovery and an individual's rehabilitation goals (Rothi & Horner, 1983) and should be clearly stated to match appropriate training outcomes.

Learning principles of training

Regardless of whether substitution or restitution is pursued, training can lead to learning and permanent improvements in motor functions (Krakauer, 2006). This learning process can be optimized by incorporating important motor learning principles, i.e. massed practice (repetitive practice with little to no rest periods in one session), appropriate dosage (sufficient frequency and duration), variable practice (practice under varied conditions) and task-specific practice (focusing on tasks directly related to the desired function) that are grounded in motor learning and neuroplasticity theory (Maier et al., 2019).

Task specific gait training strategies using technology

Various task-specific strategies for the training of functional gait ability, from no-tech to low-tech and high-tech, have dominated the rehabilitation programs in the last decades. Traditionally, therapists-supported stepping is applied in rehabilitation to restore a normal gait pattern (Werner, 2002). With the rapid development of technology, several learning principles (like massed practice, appropriate dosage and task specific practice; Maier et al., 2019) are more and more effectively incorporated into robot-assisted therapy for walking. In robot-assisted therapy, massed practice is presumably achieved through repeated gait cycles, while the dosage can be precisely controlled to ensure sufficient frequency and duration to enhance motor learning in a task-specific fashion (van Dellen & Labruyère, 2022). Technology is nowadays increasingly seen as having potential to enhance gait training, with training outcomes similar or even better than conventional treatment (Laut et al., 2016), while placing less strain on therapists (Dellen & Labruyère, 2022; Werner et al., 2002). Roughly two categories of technology used for gait training post-stroke can be distinguished. First, there are low-tech devices such as overground gait trainers that provide body weight support, like the Andago V2.0 (Hocoma AG, Volketswil, Switzerland), as well as simple systems that apply perturbations to the gait pattern, such as the C-Mill (ForceLink, Culemborg, The Netherlands). Second, there are high-tech (often costly) devices like exoskeletal-type robots that involve a more complex level of design and construction to support gait and allow to implement rehabilitation training strategies; these devices Like the Lokomat Pro version 6.0 (Hocoma AG, Volketswil, Switzerland) apply adjustable forces to each of the joints while assuring that the device matches the design of the legs and the specific anthropomorphic dimensions of the user (Laut et al., 2016). All technological rehabilitation devices can provide assistance, resistance, or a combination of both to influence the gait pattern.

Assisting or resisting gait function post stroke to explore neuromuscular and motor control

In post-stroke gait training, assistance and resistance are key concepts aimed at improving walking function. Assistance provides external support to help the patient with movements, while resistance applies external forces to challenge walking ability. These strategies are aimed at improving gait performance, though their effectiveness can vary depending on individual needs. Assisting or resisting the gait pattern enables systematic manipulation of key phases within the gait cycle. By perturbing specific aspects of gait function such as limb advancement during late stance and swing or propulsion in late stance, both common impairments post stroke (Turns et al., 2007), it becomes possible to examine how muscle activity and motor control adapt to these disturbances. Clinically, understanding these adaptation strategies can help refine rehabilitation technologies to facilitate motor learning and promote long-term functional recovery, ultimately improving mobility and independence in daily life. In this thesis immediate effects of different types of assisting or resisting technologies on muscle activation and motor control adaptations will be investigated.

Assisting the gait pattern using the Lokomat exoskeleton

The Lokomat exoskeleton

A popular example of a high-tech device for gait training that can incorporate motor learning principles like massed practice is the Lokomat. The Lokomat combines a treadmill with a body weight support system and an actuated exoskeleton that provides adjustable movement support during the entire predefined gait cycle. The Lokomat offers the opportunity to set three parameters: body weight support (0-100%), movement support (0-100% for both legs separately) and treadmill speed (0.14-0.89 m/s).

Assisting limb advancement within the Lokomat

The Lokomat enables precise regulation of the walker's movement. For post-stroke patients, this allows for the control and assistance of limb advancement. The movement support, or so called 'guidance', applies a torque when the leg deviates from the predefined gait cycle (Hussain et al., 2011; Riener et al., 2010). The magnitude of the torque is determined by the level of guidance selected (Riener et al., 2010). The option of setting guidance levels asymmetrically was incorporated into the Lokomat to specifically assist the user based on his capabilities while at the same time allowing more active contributions when possible. Previous research in able-bodied individuals showed that lower levels of symmetrical guidance coincide with higher levels of muscle activity (van Kammen et al., 2016; Moreno et al., 2013). Arguably, when setting guidance levels asymmetrically during Lokomat guided gait, higher muscle amplitudes could be evoked that matches the user' capabilities of both legs separately. This may be particularly relevant for training situations as active participation is an important prerequisite for motor learning and effective training effects to occur (Lotze et al., 2003).

The Lokomat has been commercially available since 2000 (Laszlo et al., 2023). However, at that time, the exact effects of the Lokomat on the gait pattern of able-bodied individuals and stroke patients were unknown. In the past two decades, the Lokomat has been extensively studied for its effectiveness in rehabilitation post stroke. Research has primarily focused on its ability to improve gait function (i.e. Husemann et al., 2007 and van Nunen et al., 2015) and on unraveling the effects of the adjustable parameters incorporated within the Lokomat exoskeleton (i.e. van Kammen et al., 2017). In the first two studies of this thesis, we aimed to extend beyond the existing literature by assessing the effects of assisting gait function (a)symmetrically during walking within the Lokomat exoskeleton on neuromuscular control. In **chapter 2** (able-bodied participants) and **chapter 3** (individuals post-stroke), we explore the effects of (a)symmetrical guidance provided by the Lokomat on muscle activity and its potential for training muscle activity of muscles specifically related to limb advancement and propulsion.

Resisting the gait pattern by external restraining forces

High-tech devices can influence the gait pattern in a very detailed but constrained manner, i.e. the Lokomat forces the user to walk in a predefined pattern allowing minimal variability (Hussain et al., 2011; Riener et al., 2010), which may limit motor learning. Additionally, the Lokomat may be primarily focused on restitution of function as the device forces walkers to follow the predefined pattern and the design of the exoskeleton limits certain compensation behavior (i.e. circumduction is not possible to compensate for reduced knee flexion as movements are limited to the sagittal plane). Certain less-constrained strategies may allow for more variability and compensation or adaptive behavior of the walker during training. A simple, low-tech, and affordable strategy to resist forward gait is to apply a horizontal restraining force while a person walks on a treadmill. This additional load can be introduced using a pulley system (Veeger et al., 1989), placed behind the treadmill to create a horizontal restraining force onto the trunk or leg(s) during walking, which has already shown promising improvements in propulsion. Previous research showed that restraining forces can be used to induce a higher propulsive impulse in individuals post-stroke (Lewek et al., 2018). Furthermore, the increase in propulsive impulse continued 3 minutes upon removal of the restraining force (Lewek et al., 2018) suggesting the presence of a learning effect. Although the use of this latent propulsive capacity and the presence of a learning effect seem promising for training propulsion post-stroke, it is yet unknown how this type of restraining forces could be optimized to enhance its effects. More specifically it is unknown how the immediate effect of restraining force depends on factors such as gait speed and the magnitude of the horizontal restraining force and on its point of application. Therefore, **chapter 3 and chapter 4** will explore the effects of restraining forces applied to the leg and trunk in able-bodied participants and stroke patients respectively, while walking on a motor controlled instrumented treadmill.

Aims of this thesis

The work of this thesis aims to contribute to our understanding of effective gait training the field of stroke rehabilitation. The main aims of this thesis are to (1) explore the use of different rehabilitation technologies for assisting or resisting the gait pattern to manipulate gait function (more specifically limb advancement and/or propulsion), and muscle activity related to gait function; and (2) determine the potential of these strategies for improving temporal and mechanical gait characteristics during rehabilitation post stroke. The first research question that will be addressed in **chapter 2** is if and to what extent the (a)symmetrical guidance feature incorporated into the high-tech Lokomat exoskeleton can be used to manipulate active muscular contributions for gait function in able-bodied individuals. Following this, **chapter 3** will determine the potential of (a) symmetrical guidance and its effects on muscle activity in individuals post-stroke. Chapter 4 and 5 will focus on a less-constrained system to apply restraining forces to either one leg or the trunk to target paretic propulsion. As such, **chapter 4** will map the effects of unilateral swing leg resistance on gait function during treadmill walking in able-bodied individuals. **Chapter 5** will answer the fourth and last research question of this thesis; whether a restraining force applied to the trunk is able to manipulate propulsion and other important gait characteristics during treadmill walking post-stroke. Finally, the results of the four experimental chapters, and their implication for gait rehabilitation after stroke will be discussed and summarized in **chapter 6**.

References

Balaban, B., & Tok, F. (2014). Gait disturbances in patients with stroke. *Pm&r*, *6*(7), 635-642.

Cramer, S. C., Wolf, S. L., Adams Jr, H. P., Chen, D., Dromerick, A. W., Dunning, K., ... & Broderick, J. P. (2017). Stroke recovery and rehabilitation research: issues, opportunities, and the National Institutes of Health StrokeNet. *Stroke*, *48*(3), 813-819.

GBD 2016 Lifetime Risk of Stroke Collaborators. (2018). Global, regional, and country-specific lifetime risks of stroke, 1990 and 2016. *New England Journal of Medicine*, *379*(25), 2429-2437.

GBD 2019 Stroke Collaborators. (2021). Global, regional, and national burden of stroke and its risk factors, 1990–2019: a systematic analysis for the Global Burden of Disease Study 2019. *The Lancet. Neurology*, *20*(10), 795.

van Dellen, F., & Labruyère, R. (2022). Settings matter: a scoping review on parameters in robot-assisted gait therapy identifies the importance of reporting standards. *Journal of neuroengineering and rehabilitation*, *19*(1), 40.

Duncan, P. W., Zorowitz, R., Bates, B., Choi, J. Y., Glasberg, J. J., Graham, G. D., ... & Reker, D. (2005). Management of adult stroke rehabilitation care: a clinical practice guideline. *stroke*, *36*(9), e100-e143.

Husemann, B., Muller, F., Krewer, C., Heller, S., & Koenig, E. (2007). Effects of locomotion training with assistance of a robot-driven gait orthosis in hemiparetic patients after stroke: a randomized controlled pilot study. *Stroke*, *38*(2), 349-354.

Hussain, S., Xie, S. Q., & Liu, G. (2011). Robot assisted treadmill training: mechanisms and training strategies. *Medical engineering & physics*, *33*(5), 527-533.

Hsu, A. L., Tang, P. F., & Jan, M. H. (2003). Analysis of impairments influencing gait velocity and asymmetry of hemiplegic patients after mild to moderate stroke. *Archives of physical medicine and rehabilitation*, *84*(8), 1185-1193.

Jørgensen, H. S., Nakayama, H., Raaschou, H. O., & Olsen, T. S. (1995). Recovery of walking function in stroke patients: the Copenhagen Stroke Study. *Archives of physical medicine and rehabilitation*, *76*(1), 27-32.

van Kammen, K., Boonstra, A. M., van der Woude, L. H., Reinders-Messelink, H. A., & den Otter, R. (2017). Differences in muscle activity and temporal step parameters between Lokomat guided walking and treadmill walking in post-stroke hemiparetic patients and healthy walkers. *Journal of neuroengineering and rehabilitation*, *14*, 1-11.

van Kammen, K., Boonstra, A. M., van der Woude, L. H., Reinders-Messelink, H. A., & den Otter, R. (2016). The combined effects of guidance force, bodyweight support and gait speed on muscle activity during able-bodied walking in the Lokomat. *Clinical biomechanics*, *36*, 65-73.

Kitago, T. O. M. O. K. O., & Krakauer, J. W. (2013). Motor learning principles for neurorehabilitation. *Handbook of clinical neurology*, *110*, 93-103.

Kollen, B. J., Lennon, S., Lyons, B., Wheatley-Smith, L., Scheper, M., Buurke, J. H., ... & Kwakkel, G. (2009). The effectiveness of the Bobath concept in stroke rehabilitation: what is the evidence?. *Stroke*, *40*(4), e89-e97.

Krakauer, J. W., Mazzoni, P., Ghazizadeh, A., Ravindran, R., & Shadmehr, R. (2006). Generalization of motor learning depends on the history of prior action. *PLoS biology*, *4*(10), e316.

Kwakkel, G., Wagenaar, R. C., Koelman, T. W., Lankhorst, G. J., & Koetsier, J. C. (1997). Effects of intensity of rehabilitation after stroke: a research synthesis. *Stroke*, *28*(8), 1550-1556.

Kwakkel, G., Veerbeek, J. M., van Wegen, E. E., & Wolf, S. L. (2015). Constraint-induced movement therapy after stroke. *The Lancet Neurology*, *14*(2), 224-234.

Langhorne, P., Bernhardt, J., & Kwakkel, G. (2011). Stroke rehabilitation. The Lancet, 377(9778), 1693-1702.

Lamontagne, A., Stephenson, J. L., & Fung, J. (2007). Physiological evaluation of gait disturbances post stroke. *Clinical Neurophysiology*, *118*(4), 717-729.

Laut, J., Porfiri, M., & Raghavan, P. (2016). The present and future of robotic technology in rehabilitation. *Current physical medicine and rehabilitation reports*, *4*, 312-319.

Laszlo, C., Munari, D., Maggioni, S., Knechtle, D., Wolf, P., & De Bon, D. (2023). Feasibility of an intelligent algorithm based on an assist-as-needed controller for a robot-aided gait trainer (Lokomat) in neurological disorders: a longitudinal pilot study. *Brain Sciences*, *13*(4), 612.

Lewek, M. D., Raiti, C., & Doty, A. (2018). The presence of a paretic propulsion reserve during gait in individuals following stroke. *Neurorehabilitation and neural repair*, *32*(12), 1011-1019.

Lotze, M., Braun, C., Birbaumer, N., Anders, S., & Cohen, L. G. (2003). Motor learning elicited by voluntary drive. *Brain : a journal of neurology*, *126*(Pt 4), 866–872. https://doi.org/10.1093/brain/awg079

Maulden, S. A., Gassaway, J., Horn, S. D., Smout, R. J., & DeJong, G. (2005). Timing of initiation of rehabilitation after stroke. *Archives of physical medicine and rehabilitation*, *86*(12), 34-40.

Moreno, J. C., Barroso, F., Farina, D., Gizzi, L., Santos, C., Molinari, M., & Pons, J. L. (2013). Effects of robotic guidance on the coordination of locomotion. *Journal of neuroengineering and rehabilitation*, *10*, 1-15.

Maier, M., Ballester, B. R., & Verschure, P. F. (2019). Principles of neurorehabilitation after stroke based on motor learning and brain plasticity mechanisms. Frontiers in systems neuroscience, 13, 74.

Murphy, S. J., & Werring, D. J. (2020). Stroke: causes and clinical features. Medicine, 48(9), 561-566.

van Nunen, M. P., Gerrits, K. H., Konijnenbelt, M., Janssen, T. W., & de Haan, A. (2015). Recovery of walking ability using a robotic device in subacute stroke patients: a randomized controlled study. Disability and Rehabilitation: Assistive Technology, 10(2), 141-148.

Nadeau, S., Gravel, D., Arsenault, A. B., & Bourbonnais, D. (1999). Plantarflexor weakness as a limiting factor of gait speed in stroke subjects and the compensating role of hip flexors. Clinical biomechanics, 14(2), 125-135.

Perry J., Burnfield J. M. (2010). Gait Analysis: Normal and Pathological Function, 2nd Edn. Thorofare, NJ: SLACK Incorporated.

Riener, R., Lünenburger, L., Maier, I. C., Colombo, G., & Dietz, V. (2010). Locomotor Training in Subjects with Sensori-Motor Deficits: An Overview of the Robotic Gait Orthosis Lokomat. Journal of Healthcare Engineering, 1(2), 197-216.

Rothi, L. J., & Horner, J. (1983). Restitution and substitution: Two theories of recovery with application to neurobehavioral treatment. Journal of Clinical and Experimental Neuropsychology, 5(1), 73-81.

Schmid, A., Duncan, P. W., Studenski, S., Lai, S. M., Richards, L., Perera, S., & Wu, S. S. (2007). Improvements in speed-based gait classifications are meaningful. Stroke, 38(7), 2096–2100. https://doi.org/10.1161/STROKEAHA.106.475921

Shahid, J., Kashif, A., & Shahid, M. K. (2023). A comprehensive review of physical therapy interventions for stroke rehabilitation: impairment-based approaches and functional goals. Brain Sciences, 13(5), 717.

Turns, L. J., Neptune, R. R., & Kautz, S. A. (2007). Relationships between muscle activity and anteroposterior ground reaction forces in hemiparetic walking. Archives of physical medicine and rehabilitation, 88(9), 1127-1135.

Veeger, D. H. E. J., Van der Woude, L. H., & Rozendal, R. H. (1989). The effect of rear wheel camber in manual wheelchair propulsion. J Rehabil Res Dev, 26(2), 37-46.

Werner, C., Bardeleben, A., Mauritz, K. H., Kirker, S., & Hesse, S. (2002). Treadmill training with partial body weight support and physiotherapy in stroke patients: a preliminary comparison. European Journal of Neurology, 9(6), 639-644.

Whittle, M. W. (2014). Gait analysis: an introduction. Butterworth-Heinemann.

Zorgstandaard CVA/TIA. (2012). https://www.kennisnetwerkcva.nl/wp-content/uploads/2018/09/Zorgstandaard-CVA-TIA-printversie.pdfersie.pdf

CHAPTER 2

2

The effect of asymmetric movement support on muscle activity during Lokomat guided gait in able-bodied individuals

Sylvana Weiland
Ineke H. Smit
Heleen Reinders-Messelink
Lucas H.V. van der Woude
Klaske van Kammen
Rob den Otter

PLoS One, 2018, 13(6), e0198473

Abstract

Background. To accommodate training for unilaterally affected patients (e.g. stroke), the Lokomat (a popular robotic exoskeleton-based gait trainer) provides the possibility to set the amount of movement guidance for each leg independently. Given the interlimb couplings, such asymmetrical settings may result in complex effects, in which ipsilateral activity co-depends on the amount of guidance offered to the contralateral leg. To test this idea, the effect of asymmetrical guidance on muscle activity was explored.

Methods. 15 healthy participants walked in the Lokomat at two speeds (1 and 2 km/h) and guidance levels (30% and 100%), during symmetrical (both legs receiving 30% or 100% guidance) and asymmetrical conditions (one leg receiving 30% and the other 100% guidance) resulting in eight unique conditions. Activity of the right leg was recorded from Erector Spinae, Gluteus Medius, Biceps Femoris, Semitendinosus, Vastus Medialis, Rectus Femoris, Medial Gastrocnemius and Tibialis Anterior. Statistical Parametric Mapping was used to assess whether ipsilateral muscle activity depended on guidance settings for the contralateral leg.

Results. Muscle output amplitude not only depended on ipsilateral guidance settings, but also on the amount of guidance provided to the contralateral leg. More specifically, when the contralateral leg received less guidance, ipsilateral activity of Gluteus Medius and Medial Gastrocnemius increased during stance. Conversely, when the contralateral leg received more guidance, ipsilateral muscle activity for these muscles decreased. These effects were specifically observed at 1 km/h, but not at 2 km/h.

Conclusions. This is the first study of asymmetrical guidance on muscle activity in the Lokomat, which shows that ipsilateral activity co-depends on the amount of contralateral guidance. In therapy, these properties may be exploited e.g. to promote active contributions by the more affected leg. Therefore, the present results urge further research on the use of asymmetrical guidance in patient groups targeted by Lokomat training.

Introduction

For persons with motor impairments due to neurological disease or trauma, the ability to walk is an important determinant of independent functioning [1]. Clinical evidence shows that the major determinants of functional outcome are task specificity and training intensity: in order to re-learn to walk, patients have to be involved in walking activity and produce a large number of stepping movements [2,3]. In the past decade, robotic gait trainers have been developed that eliminate the need for manual assistance by therapist, as actuated exoskeletons allow automated support during more stepping movements. As studies on the clinical effectiveness of robot-assisted gait trainers (RAGT) thus far have yielded positive (see e.g. [4]) as well as negative results (see e.g. [5]), there is a need to more fully understand how training can be shaped through parameter settings (e.g. the amount of guidance), and to explore the specific possibilities offered by RAGT.

RAGTs have the unique ability to provide automated movement support (so-called 'guidance'), and guide the legs through a predefined gait pattern [6]. A popular treadmill-based robotic device for gait training is the Lokomat [6], which is able to offer guidance based on two possible strategies: path control and impedance control [7,8]. In this study, the impedance control strategy was used, which allows adjustable control over the amount of supportive force that 'guides' the legs through the gait cycle [8]. A significant proportion of the target population for RAGT (e.g. stroke, cerebral palsy) suffers from asymmetrical locomotor impairments, and would therefore require more assistance for one leg than for the other. To accommodate these training needs, guidance levels in the Lokomat can be set separately for each limb, making it possible to tune the required contributions specifically to the capacity of each leg. Although the neuromuscular control of Lokomat guided gait [9-11] and the effects of symmetrical guidance [12,13] have been described in good detail, it is still unclear how the muscle activity that drives Lokomat guided gait is affected when guidance is provided asymmetrically.

Arguably, under asymmetrical conditions, the involvement of muscles may not only depend on the guidance offered to the ipsilateral leg, but also on the amount of guidance provided to the contralateral leg. Stable bipedal progression is organized bilaterally, and depends on the coupled control of the two legs. Studies in which unequal contributions of the legs are required show that the phasing and amplitude of ipsilateral muscle activity depend on the activity of the contralateral leg [14-20]. These couplings likely involve networks that integrate ascending (e.g. hip afferents and load receptors) and descending (spinal and supra-spinal) commands [21]. The bilateral involvement of these networks can be demonstrated quite dramatically in tasks where movements of one of the legs are blocked. Under these conditions, structurally phased activity in the passive leg can still be induced through cyclical, task specific movements of the contralateral leg [17,18,21-23]. Observations like these suggest that levels of gait specific neuromuscular activity in the ipsilateral leg may be tuned to the contributions made by the contralateral leg.

Previous research on Lokomat guided gait has shown that the contribution of muscles is inversely related to the amount of symmetrical guidance that is provided, so that higher levels of guidance generally induce lower levels of muscle activity [12,13]. It can be argued that, when guidance is provided asymmetrically, varying the level of contralateral guidance may alter the activity of ipsilateral muscles. For therapists, this may open up new opportunities to stimulate more active participation of patients during training. To test these ideas in the present study, able-bodied subjects walked in the Lokomat under symmetrical as well as asymmetrical guidance conditions. The aim was to assess if and to what extent levels of ipsilateral muscle activity depend on the level of guidance provided to the contralateral leg. As previous work has shown that the effects of guidance depend on gait speed [12], a secondary aim was to assess if the effects of asymmetrical guidance depend on treadmill speed. To this end, muscle activity was assessed in healthy participants during symmetrical and asymmetrical guidance conditions at both 1 and 2 km/h.

Methods

Participants

Fifteen healthy adults (5 male, 22.2 ± 2.1 year, 1.73 ± 0.07 meter and 65.6 ± 9.52 kg) participated in this study. None of the subjects had a condition that is known to affect the gait pattern or muscle activity. All experimental procedures were approved by the Ethical Committee of the Center for Human Movement Sciences (University Medical Center Groningen, the Netherlands) and performed according to the declaration of Helsinki for medical research involving human subjects [24]. All subjects provided written informed consent prior to the procedure.

Materials

Participants walked in the LokomatPro V6.0 (Hocoma AG, Volketswil, Switzerland) at the rehabilitation center 'Revalidatie Friesland' in Beetsterzwaag, the Netherlands. The Lokomat is a stationary gait orthosis that moves the legs through the gait cycle by means of actuated knee and hip joints that are integrated into the exoskeleton [8]. The Lokomat offers the opportunity to set the level of guidance, body weight support (BWS) and treadmill speed. In the present study, guidance was provided by means of an impedance controller, which applies a controllable torque to each joint to keep the legs on the reference trajectory [7,8]. When guidance is reduced, a smaller torque that pushes the legs towards the pre-defined trajectory is applied compared to maximal guidance [8].

Surface electromyography (EMG) was used to record muscle activation patterns from eight muscles of the right (ipsilateral) leg: Erector Spinae (ES), Gluteus Medius (GM), Biceps Femoris (BF), Semitendinosus (ST), Vastus Medialis (VM), Rectus Femoris (RF), Medial Gastrocnemius (MG) and Tibialis Anterior (TA). Ag/AgCl electrodes (Kendall/Tyco ARBO; Warren, MI, USA) with a 10 mm diameter and a minimum electrode distance of 25 mm, were used and placed following the SENIAM recommendations [26]. To ensure good conduction, body hair and dead skin cells were removed by shaving and abrading the electrode sites, whereupon these sites were cleaned with alcohol. EMG signals were pre-amplified and A/D converted (22 bits) using a 32-channel Porti7 portable recording system (Twente Medical Systems, Enschede, The Netherlands). The recording system has a common mode rejection of 0.90 dB, a 2 mVpp noise level and an input impedance of 0.10 GV. To detect gait cycle events, custom made insoles (Pedag international VIVA, Berlin, Germany) were used containing one pressure sensor under the heel and three pressure sensors under the forefoot (FSR402, diameter 18 mm, loading 10–1000 g). The EMG and pressure sensor data were simultaneously sampled via the Porti7 system at 2048 Hz before storage on a computer for offline processing, so that no synchronization was required.

Procedure

Participants walked sixteen trials in the Lokomat without BWS, and without foot straps, while guidance (30% and 100%) and treadmill speed (1 and 2 km/h) were varied. This resulted in symmetrical trials where both legs received 30% guidance (30 ipsilateral (IL) / 30 contralateral (CL)) or 100% guidance (100IL/100CL) and asymmetrical trials where contralateral guidance was higher (30IL/100CL) or lower (100IL/30CL) compared to ipsilateral guidance at both speed levels (see Table 1). To prevent order effects, a randomized order of the eight unique conditions (30IL/30CL, 100IL/100CL, 30IL/100CL and 100IL/30CL, at both 1 and 2 km/h) was presented first (block 1). Subsequently, the same conditions were presented a second time in a counter-balanced order (block 2) to control for learning and fatigue. To ensure that an approximately equal number of steps was measured, trials at 1 km/h lasted 120 seconds and trials at 2 km/h 60 seconds. The quality of the EMG and pressure sensor signals was constantly monitored during the experiment. Before starting each trial, an acclimatization period of two minutes was provided to the participants to get accustomed to the specific Lokomat settings.

Table 1. The conditions offered to the participants twice.

	30% guidance ipsilateral		100% guidance ipsilateral	
	Symmetrical condition	Asymmetrical condition	Symmetrical condition	Asymmetrical condition
1 km/h	*30IL/30CL*	*30IL/100CL*	*100IL/100CL*	*100IL/30CL*
2 km/h	*30IL/30CL*	*30IL/100CL*	*100IL/100CL*	*100IL/30CL*

Both legs received 30% guidance (30 ipsilateral (IL) / 30 contralateral (CL)) or 100% guidance (100IL/100CL) during symmetrical trials and a higher (30IL/100CL) or lower (100IL/30CL) level of contralateral guidance during asymmetrical trials.

Data analysis

EMG and pressure sensor data were analysed offline using custom-made software routines in Matlab (Matlab 2015a; Mathworks, Natick, MA). The EMG data were filtered using a 10 Hz fourth order Butterworth high-pass filter to attenuate movement artefacts. Subsequently, EMG data were full wave rectified, low-pass filtered using a 20 Hz fourth order Butterworth filter, time normalized (from heelstrike to heelstrike based on insole pressure data) to 100 data points per gait cycle, and amplitude normalized with respect to the percentage of the maximal amplitude observed over all of the trials, for each muscle and each participant separately. For the analysis, muscle activity of the corresponding trials performed in block 1 and 2 were averaged.

Statistical analysis

One approach to the statistical analysis of dynamic EMG is to compare the (summed or averaged) amplitudes within pre-specified time windows, e.g. the single stance phase, the swing phase and the first and second double support phase [12], or within specific time windows that are set arbitrarily based on the observed profiles [11]. However, previous research on muscle activity in the Lokomat has shown that activity may occur within windows that are a-typical for unrestrained overground or treadmill walking [11]. As this makes it difficult to define time windows a priori, we choose to assess the averaged and time-normalized EMG time series using Statistical Parametric Mapping (SPM).

SPM repeated measures ANOVA's (RM) were used to assess the effects of the factors Guidance (30 and 100%) and Symmetry (asymmetrical and symmetrical) for both speed levels, and all eight muscles, separately. SPM was originally developed to deal with the spatial correlation inherent in neuroimaging data [27], and has quite recently also been applied to biomechanical data (e.g. [28]). The output of an SPM RM analysis is a 'Statistical Parametric Map' that consists of F-values (SPM{F}) calculated separately for each data point *t* in the time normalized gait cycle (t = 1...100). This implies that no time windows need to be defined a priori. Since there is a spatial correlation present in the smoothed EMG signal, the number of independent tests does not equal the number of data points in the time normalized gait cycle. To maintain the familywise error rate at the intended 5%, SPM estimates the trajectory smoothness using temporal gradients [29]. Based on these smoothness estimates and the number of data points, the actual number of independent tests can be determined. The required thresholds are then calculated based on an a priori alpha value (0.05 in this case) using Random Field Theory (RFT) expectations with regard to the field-wide maximum [30].

F-values that exceed the calculated threshold often occur in clusters of data points, representing significant segments of the time series in which less than 5% of smooth random curves would be expected to cross. The exact probability that a cluster of a specific size crosses the threshold can be calculated for each of these supra-threshold clusters, based on cluster size and RFT distribution(s) for SPM{F} topology. It is important to note here that SPM eliminates the need for a priori decisions with regard to regions of interest, and thus avoids potential bias due to e.g. data driven window setting. For a more in-depth discussion of SPM and its applications, we refer to more elaborate texts on this topic [27-29].

To be able to answer the research question, the effects of contralateral guidance on ipsilateral muscle activity (referred to as the asymmetrical guidance effects) were assessed by interpreting the interaction effects of Guidance by Symmetry. In addition, assessment of the main effects of Guidance (referred to as the general effects of guidance) allowed us to establish if and how possible effects of asymmetrical guidance are related to guidance induced inhibitions/facilitations of ipsilateral activity. Supra-threshold clusters for the main effect of Guidance indicate a difference in the amplitude normalized, ipsilateral muscle activity between 30 and 100% guidance conditions. Supra-threshold clusters for Guidance by Symmetry interaction indicate that the effects of guidance on the amplitude-normalized EMG values in the ipsilateral leg, depend on the level of guidance provided to the contralateral leg. In case of supra-threshold clusters, the direction of the effects was determined by comparing the condition means within the cluster. All SPM analyses were done using open-source spm1d code (v.M0.1, www.spm1d.org) in Matlab (R2016a, The Mathworks Inc, Natick, MA).

Results

General effects of guidance on muscle activity

Significant main effects of Guidance were found at 1 km/h, indicating that the level of ipsilateral muscle activity was negatively associated with the level of ipsilateral guidance that was provided. Muscle activity in the ipsilateral leg decreased when guidance increased from 30% to 100% in four supra-threshold clusters in ES (51.8-52.3%, p=0.0496; 53.4-54.6%, p=0.0477; 58.2-63.0%, p=0.0256; 63.0-70.7% , p=0.0084; critical threshold (CT) = 13.5 for all clusters) two supra-threshold clusters in BF (61.8-62.3%, p=0.0495; 63.8-65.8%, p=0.0407; CT=14.0), two supra-threshold clusters in ST (13.8-14.1%, p=0.0497; 17.7-18.4%, p=0.0490; CT=14.1), and one supra-threshold cluster in RF (79.6-85.6%, p=0.0152; CT=13.6). Similar main effects of guidance were observed at 2 km/h for ST as indicated by one supra-threshold cluster (96.0-96.0%, p=0.0500; CT=14.4). No other significant main effects of guidance were observed at 2 km/h.

Effects of the Guidance by Symmetry interaction on muscle activity

Results showed that at 1 km/h ipsilateral muscle activity of GM and MG depended on the level of guidance provided to the contralateral leg. Fig 1 presents EMG profiles for MG of a single representative participant at 1 km/h to illustrate the effects. As presented in this figure, the amount of ipsilateral muscle activity depends on the amount of contralateral guidance. When ipsilateral guidance was set to 30%, the amplitude of muscle activity decreased when the contralateral leg was provided 100% guidance (30IL/100CL), compared to the symmetrical 30IL/30CL condition (see fig. 1B). An opposite effect was observed when ipsilateral guidance was set to 100%, as ipsilateral activity increased when contralateral guidance was lower (100IL/30CL) compared to the symmetrical 100IL/100CL condition (see fig. 1B).

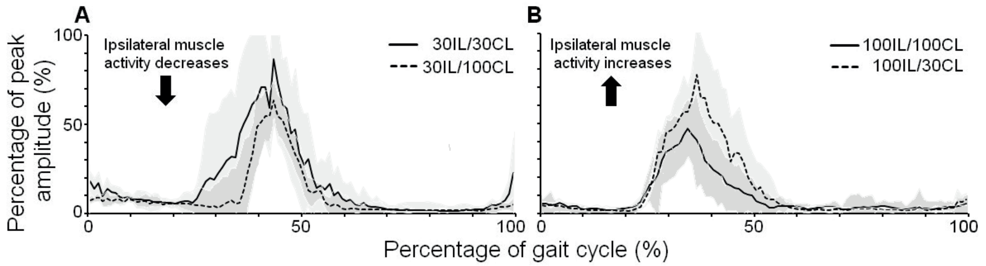

Fig 1. EMG profiles for Medial Gastrocnemius (MG) when guidance offered to the contralateral leg (CL) increases (A) or decreases (B) while guidance of the ipsilateral leg (IL) was held constant at 30% (A) or 100% (B). Time and amplitude normalized EMG profiles during walking in the Lokomat exoskeleton for a representative participant at 1 km/h. EMG amplitude is expressed as a percentage of peak amplitude recorded over all conditions. The standard deviation of the EMG profiles for each condition is presented by the dark shade.

Group averaged EMG profiles are presented together with the SPM results of the interaction effect of Guidance by Symmetry in figs 2 (ES, GM, BF and ST) and 3 (VM, RF, MG and TA). All supra-threshold clusters indicate segments of the EMG time series in which less than 5% of smooth random curves would be expected to cross the threshold (df=1,14). As becomes clear from these figures, at 1 km/h one supra-threshold cluster (38.8-49.5% of the gait cycle, CT=13,8; p=0.0008) was identified for the Guidance by Symmetry interaction in MG. Comparison of the group-averaged values within this cluster showed that when contralateral guidance was higher (30IL/100CL), ipsilateral activity decreased compared to when both legs received low levels of guidance (30IL/30CL). In contrast (see fig. 2), when contralateral guidance was lower (100IL/30CL), ipsilateral activity increased relative to when both legs received maximal guidance (100IL/100CL). Similar effects of asymmetrical guidance were present in GM at 1 km/h as indicated by five supra-threshold clusters (31.9-32.2%, p=0.0499; 33.5-39.0% p=0.0191; 43.5-44.4%, p=0.0488; 45.4-46.5%, p=0.0482; 50.6-51.6% p=0.0486, CT=13.6 for all clusters).

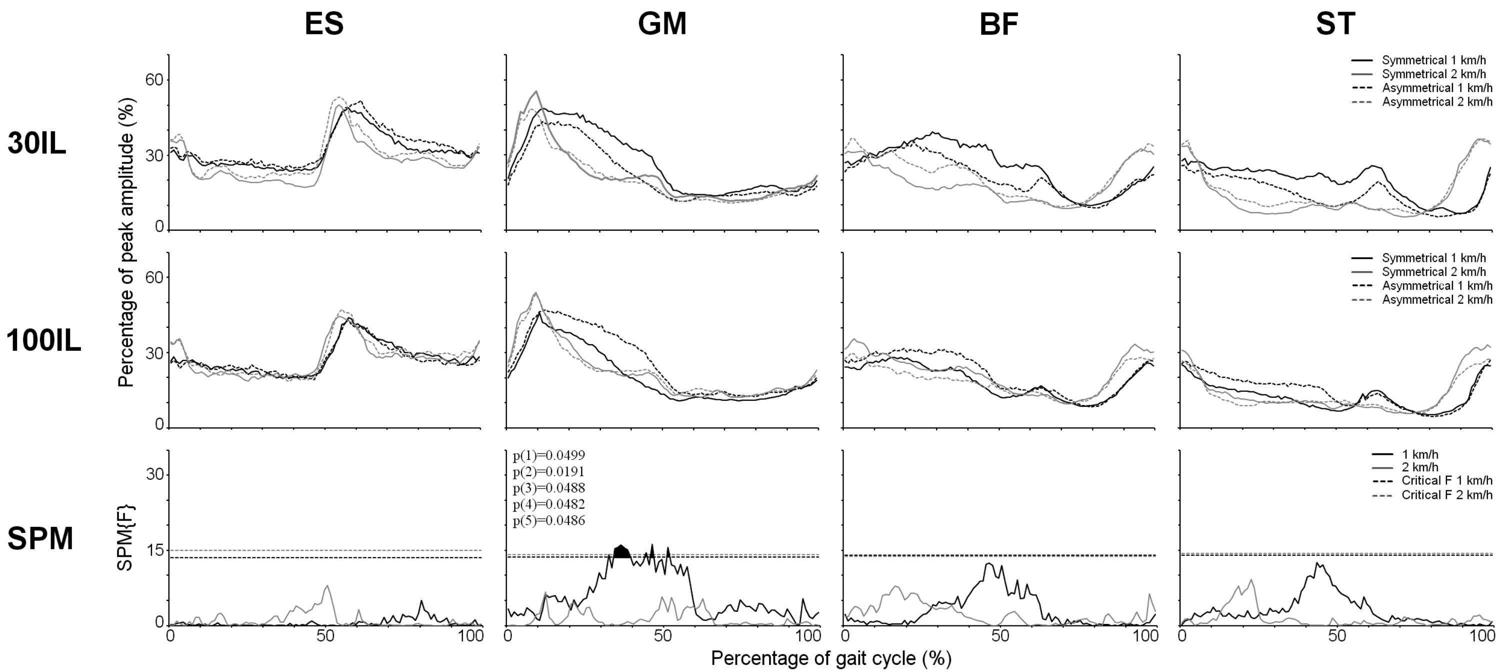

Fig 2. Average EMG profiles for Erector Spinae (ES), Gluteus Medius (GM), Biceps Femoris (BF) and Semitendinosus (ST) combined with the SPM results of the SPM analyses for the Guidance by Symmetry interaction. Averaged EMG profiles are presented for Erector Spinae (ES), Gluteus Medius (GM), Biceps Femoris (BF) and Semitendinosus (ST) combined with the results of the SPM analyses for the Guidance by Symmetry interaction. The EMG profiles are time and amplitude normalized during walking in the Lokomat exoskeleton with 30% guidance provided to the ipsilateral leg (first row) and 100% guidance provided to the ipsilateral leg (second row) at 1 km/h (black lines) and 2 km/h (grey lines) for both symmetrical (solid lines) and asymmetrical (dashed lines) conditions. EMG amplitude is expressed as a percentage of peak amplitude recorded over all conditions. The third row of figures display the SPM{F} values of the Guidance by Symmetry interaction at 1 km/h (black lines) and 2 km/h (grey lines). Threshold F values are indicated by dashed black lines (1 km/h) and dashed grey lines (2 km/h). The displayed p-values represent the chance that a supra-threshold cluster of the given cluster size would be observed in random samplings.

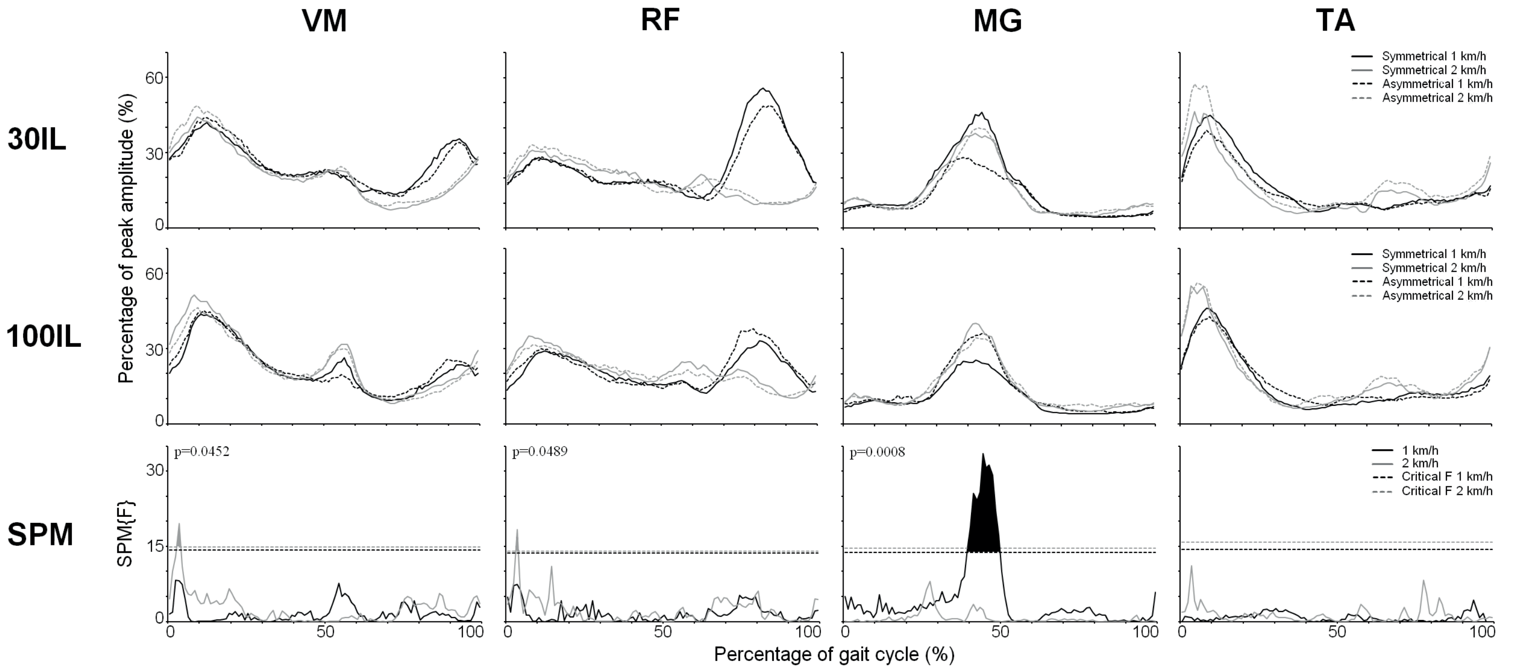

Fig 3. Average EMG profiles for Vastus Medialis (VM), Rectus Femoris (RF), Medial Gastrocnemius (MG) and Tibialis Anterior (TA) combined with the SPM results of the SPM analyses for the Guidance by Symmetry interaction. Averaged EMG profiles are presented for Vastus Medialis (VM), Rectus Femoris (RF), Medial Gastrocnemius (MG) and Tibialis Anterior (TA) combined with the results of the SPM analyses for the Guidance by Symmetry interaction. The EMG profiles are time and amplitude normalized during walking in the Lokomat exoskeleton with 30% guidance provided to the ipsilateral leg (first row) and 100% guidance provided to the ipsilateral leg (second row) at 1 km/h (black lines) and 2 km/h (grey lines) for both symmetrical (solid lines) and asymmetrical (dashed lines) conditions. EMG amplitude is expressed as a percentage of peak amplitude recorded over all conditions. The third row of figures display the SPM{F} values of the Guidance by Symmetry interaction at 1 km/h (black lines) and 2 km/h (grey lines). Threshold F values are indicated by dashed black lines (1 km/h) and dashed grey lines (2 km/h). The displayed p-values represent the chance that a supra-threshold cluster of the given cluster size would be observed in random samplings.

2

The findings of the Guidance by Symmetry interaction at 2 km/h in VM and RF were at odds with the results at 1 km/h (see fig. 3). Short supra-threshold clusters for the Guidance by Symmetry interaction were found in VM (2.3-3.6%; CT=14.0, p=0.0452) and RF (2.6-3.4% CT=14.0, p=0.0489). However, the nature of these effects was different from those observed at 1 km/h, as ipsilateral muscle amplitudes decreased when the contralateral leg received less guidance (100IL/30CL vs 100IL/100CL).

Discussion

The present study addressed the effects of contralateral guidance levels on ipsilateral muscle activity provided by the Lokomat exoskeleton in a group of able-bodied walkers. The aim was to determine if and to what extent ipsilateral muscle activity depends on the level of robotic guidance provided to the contralateral leg. The results show that, when ipsilateral guidance was held constant, ipsilateral muscle activity in MG and GM at 1 km/h was inversely related to the amount of guidance offered to the contralateral leg. These effects were observed in selected muscles at 1 km/h, but not at 2 km/h. The here established short-term effects demonstrate that during walking in the Lokomat exoskeleton, the activity of specific muscles in the ipsilateral leg can be affected by the sensorimotor state of the contralateral leg. The results urge further research on the long-term effects and the use of asymmetrical movement guidance in patient groups targeted for robot assisted gait training.

Robotic gait trainers like the Lokomat offer an alternative for manual assistance through therapists by using actuated exoskeletons to guide stepping, so as to reduce the amount of active muscular contributions. Conform this notion, and in accordance with previous work [12], at 1 km/h the amplitude of muscle activity decreased with a general increase of guidance in 4 of the 8 muscles (as indicated by the main effects of Guidance). At 2 km/h, no systematic general effects of guidance were observed, confirming earlier research on Lokomat guided walking [12] showing that effects of guidance are strongly attenuated at higher treadmill speeds. Crucial to the aims of our study, at 1 km/h, the level of ipsilateral activity in GM and MG depended on the amount of guidance that was provided to the contralateral leg. More specifically, when the ipsilateral leg received 100% guidance, activity in the muscles increased when contralateral guidance was set to 30%, compared to the symmetrical situation (100IL/100CL). Conversely, when the ipsilateral leg was provided 30% guidance, activity was reduced when 100% guidance was provided to the contralateral leg, relative to the symmetrical situation (30IL/30CL). For very short epochs (approximately 1% of the gait cycle), opposite effects were found at 2 km/h in VM and RF: lower and higher levels of contralateral guidance resulted in an inhibition and facilitation, respectively, of activity in the ipsilateral leg. Taken together, these results show that tuning the level of contralateral guidance can influence the amount of ipsilateral muscular effort.

The bilateral couplings involved in human walking become particularly evident when different task constraints are imposed on each of the individual legs, e.g. when more muscle activity is required from one leg than from the other leg (see e.g. [17,18,21-23]). Here, we imposed such an asymmetry by setting the amount of robotic guidance for each of the legs independently. With regard to the functional significance of the observed bilateral effects, it is important to note that the effects at 1 km/h occurred within time regions roughly corresponding to ipsilateral single support, suggesting that the observed inhibition and facilitation of activity was coupled to the muscular efforts to control the contralateral swing leg. During unrestrained (overground or treadmill) walking, swing leg motion is primarily regulated through its passive dynamics, reducing the need for active muscular control [31]. However, as the compensatory torques generated by the Lokomat actuators do not fully eliminate interaction torques due to exoskeleton inertia, friction, and gravity [32], this may necessitate a more active control of the swing limb during Lokomat guided walking, in particular at low guidance settings [9,11]. The data seem to confirm this, as prominent RF activity (which serves to flex the hip and extend the knee) was observed during the ipsilateral swing phase selectively at 1 km/h, and this activity was significantly attenuated when more guidance was provided in general. In addition, as MG is known to control contralateral step length [33], an increase in ipsilateral MG activity during 100IL/30CL compared to 100IL/100CL may be facilitated by the requirement of more active control of the swing limb to overcome inertia. Therefore, it cannot be excluded that the facilitatory effects of lowered contralateral guidance in GM and MG are an indirect result of uncompensated exoskeleton inertia when little movement support is provided.

The functional interpretation of the effects observed at 2 km/h (i.e. a facilitation of early stance activity in VM and RF as a result of lowered contralateral guidance) is not straightforward. Because, consistent with the literature [12], no main effects were detected for Guidance at 2 km/h in these muscles, it can be excluded that these effects were directly related to the imposed guidance manipulations. An explanation why no facilitatory effects of lowered contralateral guidance were found at 2 km/h might be related to attenuated exoskeleton inertia at 2 km/h. Since no consistent effect of guidance is present at 2 km/h, it seems likely that swing control at higher speed levels is less affected by the inertia of the exoskeleton and relies more on passive dynamics like in unrestrained walking. In this situation, the imposed asymmetry of guidance does not affect ipsilateral leg control via bilateral couplings. Lastly, it is important to note that, although the effects of contralateral guidance on the ipsilateral leg were consistent between participants, their magnitude was small and they appeared only briefly (see figure 3).

Although the here reported effects of asymmetrical guidance were quite subtle, the observed facilitatory effects may stimulate active contributions by patients during robot assisted gait. Active participation of trainees is an important prerequisite for motor learning and effective training to occur [34,35]. Because robotic guidance is known to reduce the need for muscular effort [12,13], the here reported facilitatory effects of asymmetrical guidance may be particularly desirable. Particularly when patients with unilateral gait disturbances require full robotic guidance to support their affected leg, the required neuromuscular activity necessary for effective training may be stimulated by reducing the guidance level for the unaffected side. Arguably, robotic gait trainers with more advanced control strategies that allow control of the resistance of exoskeleton segments (for example exoskeletons based on force field control see e.g. [36]) may offer possibilities to produce more dramatic effects of asymmetrical support settings.

Generalization of the present results to clinical training situations, e.g. for patients with asymmetrical gait patterns, is not self-evident. First, there are indications that in patients with supraspinal lesions, asymmetrical activation of the legs may result in abnormally coordinated and exaggerated activations, due to diminished supraspinal inhibitory control over spinal pattern generating networks [17,18,37]. Second, the facilitating effects of lowered contralateral guidance were only observed at 1 km/h and not at 2 km/h, indicating that specific speed settings may be required during clinical training situations for a training effect to occur. Thirdly, in the present protocol, no foot straps or BWS were provided. However, in clinical practice, such measures are often applied to support foot lift and weight bearing for patients. As both foot straps and BWS may affect neuromuscular task demands (e.g. BWS is known to generally attenuate muscle activity during Lokomat guided walking [11,12]), generalization of the here observed effects to clinical situations is not straightforward, and requires further inquiry. Lastly, inspection of the present results shows that (facilitatory or inhibitory) effects of asymmetrical guidance during Lokomat walking, may occur in regions of the gait cycle where no activity is found during unrestrained walking. Arguably, these bilateral effects may elicit inappropriate and confusing afferent and/or efferent information, and could potentially induce functionally inappropriate training stimuli. However, it must also be noted that the effects of asymmetrical guidance could be prominently and reliably induced during the stance phase in MG. Training the plantarflexors is important in e.g. stroke, as weakness in this muscle group has been shown to be a main limiting factor for functional gait performance [38,39]. Taken together, the present results should encourage further research on the use of asymmetrical guidance in robotic exoskeletons in patient groups.

Conclusions

To the best of our knowledge, this is the first study to assess the effect of asymmetrical robotic guidance in able-bodied gait. The results demonstrate that muscle output can be selectively facilitated or inhibited by tuning the guidance provided to the contralateral leg at specific speed settings. More research is needed to establish if asymmetrical guidance settings for robotic exoskeletons may be a useful addition to clinical gait training.

Acknowledgements: The authors wish to thank Emyl Smid for his technical help and advice during the study.

References

1. Bohannon RW. Selected determinants of ambulatory capacity in patients with hemiplegia. Clinical Rehabilitation. 1989;3(1):47-53.
2. Kwakkel G, Kollen B, Lindeman E. Understanding the pattern of functional recovery after stroke: facts and theories. Restorative neurology and neuroscience. 2004;22(3-5):281-299.
3. Teasell RW, Foley NC, Bhogal SK, Speechley MR. An evidence-based review of stroke rehabilitation. Topics in stroke Rehabilitation. 2003;10(1):29-58.
4. Cho JE, Yoo JS, Kim KE, Cho ST, Jang WS, Cho KH, et al. Systematic review of appropriate robotic intervention for gait functionin subacute stroke patients. BioMed research international. 2018.
5. Lefmann S, Russo R, Hillier S. The effectiveness of robotic-assisted gait training for paediatric gait disorders: systematic review. Journal of neuroengineering and rehabilitation. 2017;14(1):1.
6. Calabrò RS, Cacciola A, Bertè F, Manuli A, Leo A, Bramanti A, et al. Robotic gait rehabilitation and substitution devices in neurological disorders: where are we now?. Neurological Sciences. 2016;37(4):503-14.
7. Hussain S, Xie SQ, Liu G. Robot assisted treadmill training: Mechanisms and training strategies. Medical engineering & physics. 2011;33(5):527-33.
8. Riener R, Lünenburger L, Maier IC, Colombo G, Dietz V. Locomotor training in subjects with sensori-motor deficits: an overview of the robotic gait orthosis lokomat. Journal of Healthcare Engineering. 2010;1(2):197-216.
9. Hidler JM, Wall AE. Alterations in muscle activation patterns during robotic-assisted walking. Clinical Biomechanics. 2005;20(2):184-93.
10. Gizzi L, Nielsen JF, Felici F, Moreno JC, Pons JL, Farina D. Motor modules in robot-aided walking. Journal of neuroengineering and rehabilitation. 2012;9(1):76.
11. Van Kammen K, Boonstra A, Reinders-Messelink H, den Otter R. The combined effects of body weight support and gait speed on gait related muscle activity: a comparison between walking in the Lokomat exoskeleton and regular treadmill walking. PloS one. 2014;9(9):e107323.
12. van Kammen K, Boonstra AM, van der Woude LH, Reinders-Messelink HA, den Otter R. The combined effects of guidance force, bodyweight support and gait speed on muscle activity during able-bodied walking in the Lokomat. Clinical biomechanics. 2016;36:65-73.
13. Moreno JC, Barroso F, Farina D, Gizzi L, Santos C, Molinari M, et al. Effects of robotic guidance on the coordination of locomotion. Journal of neuroengineering and rehabilitation. 2013;10(1):79.
14. Arya KN, Pandian S. Interlimb neural coupling: Implications for poststroke hemiparesis. Annals of physical and rehabilitation medicine. 2014;57(9):696-713.
15. Dietz VO. Human neuronal control of automatic functional movements: interaction between central programs and afferent input. Physiological reviews. 1992;72(1):33-69.
16. Dietz V, Michel J. Human bipeds use quadrupedal coordination during locomotion. Annals of the New York Academy of Sciences. 2009;1164(1):97-103.
17. Kautz SA, Patten C. Interlimb influences on paretic leg function in poststroke hemiparesis. Journal of neurophysiology. 2005;93(5):2460-73.

18. Kautz SA, Patten C, Neptune RR. Does unilateral pedaling activate a rhythmic locomotor pattern in the nonpedaling leg in post-stroke hemiparesis?. Journal of neurophysiology. 2006;95(5):3154-63.
19. Sherrington C. The integrative action of the nervous system. CUP Archive; 1910.
20. Swinnen S, Duysens J, editors. Neuro-behavioral determinants of interlimb coordination: a multidisciplinary approach. Springer Science & Business Media; 2012.
21. Dietz V, Müller R, Colombo G. Locomotor activity in spinal man: significance of afferent input from joint and load receptors. Brain. 2002;125(12):2626-34.
22. Ferris DP, Gordon KE, Beres-Jones JA, Harkema SJ. Muscle activation during unilateral stepping occurs in the nonstepping limb of humans with clinically complete spinal cord injury. Spinal cord. 2004;42(1):14-23.
23. Kawashima N, Nozaki D, Abe MO, Akai M, Nakazawa K. Alternate leg movement amplifies locomotor-like muscle activity in spinal cord injured persons. Journal of neurophysiology. 2005;93(2):777-85.
24. World Medical Association. World Medical Association Declaration of Helsinki. Ethical principles for medical research involving human subjects. Bulletin of the World Health Organization. 2001;79(4):373.
25. Calabrò RS, De Luca R, Leo A, Balletta T, Marra A, Bramanti P. Lokomat training in vascular dementia: motor improvement and beyond!. Aging clinical and experimental research. 2015;27(6):935-7.
26. Freriks B, Hermens H, Disselhorst-Klug C, Rau G. The recommendations for sensor and sensor placement procedures for surface electromyography. In: Hermens H, editors. European recommendations for surface electromyography. Enschede: Roessingh Research and Development; 1995. pp. 15-53.
27. Friston KJ, Holmes AP, Worsley KJ, Poline JB, Frith CD, Frackowiak RSJ. Statistical parametric maps in functional imaging: a general linear approach. Human Brain Mapping. 1995;2:189–210.
28. Pataky TC. Generalized n-dimensional biomechanical field analysis using statistical parametric mapping. Journal of Biomechanics. 2010;43:1976-1982.
29. Friston KJ, Ashburner JT, Kiebel SJ, Nichols TE, Penny WD. Statistical parametric mapping: the analysis of functional brain images. London: Elsevier; 2007.
30. Adler RJ, Taylor JE. Random fields and geometry. Springer Science & Business Media; 2009.
31. Kuo AD. A simple model of bipedal walking predicts the preferred speed–step length relationship. Journal of biomechanical engineering. 2001;123(3):264-269.
32. Duschau-Wicke A, von Zitzewitz J, Caprez A, Lunenburger L, Riener R. Path control: a method for patient-cooperative robot-aided gait rehabilitation. IEEE Transactions on Neural Systems and Rehabilitation Engineering. 2010;18(1):38-48.
33. Honeine JL, Schieppati M, Gagey O, Do MC. The functional role of the triceps surae muscle during human locomotion. PloS one. 2013;8(1).
34. Liu J, Cramer SC, Reinkensmeyer DJ. Learning to perform a new movement with robotic assistance: comparison of haptic guidance and visual demonstration. Journal of neuroengineering and rehabilitation. 2006;3(1):20.
35. Lotze M, Braun C, Birbaumer N, Anders S, Cohen LG. Motor learning elicited by voluntary drive. Brain. 2003;126(4):866-72.

2

36. Banala SK, Agrawal SK, Scholz JP. Active Leg Exoskeleton (ALEX) for gait rehabilitation of motor-impaired patients. In Rehabilitation Robotics, 2007. ICORR 2007. IEEE 10th International Conference on 2007 (pp. 401-407). IEEE.
37. Lazarus JA. Associated movement in hemiplegia: the effects of force exerted, limb usage and inhibitory training. Archives of physical medicine and rehabilitation. 1992;73(11):1044-9.
38. Olney SJ, Griffin MP, McBride ID. Temporal, kinematic, and kinetic variables related to gait speed in subjects with hemiplegia: a regression approach. Physical therapy. 1994;74(9):872-85.
39. Nadeau S, Gravel D, Arsenault AB, Bourbonnais D. Plantarflexor weakness as a limiting factor of gait speed in stroke subjects and the compensating role of hip flexors. Clinical Biomechanics. 1999;14(2):125-3.

Supporting information

S1 Data. Data of the group averaged EMG profiles and the F values calculated with SPM. The supplementary file contains a separate worksheet for eight muscles: Erector Spinae (ES), Gluteus Medius (GM), Biceps Femoris (BF), Semitendinosus (ST), Vastus Medialis (VM), Rectus Femoris (RF), Medial Gastrocnemius (MG) and Tibialis Anterior (TA). The worksheets display the group-averaged EMG profiles for each symmetrical (30IL/30CL and 100IL/100CL) and each asymmetrical (30IL/100CL and 100IL/30CL) guidance condition together with the SPM{F} values calculated for the Guidance by Symmetry interaction.
https://ndownloader.figstatic.com/files/11828456

CHAPTER 3

Effects of asymmetrical support on lower limb muscle activity during Lokomat guided gait in persons with a chronic stroke: an explorative study

Sylvana Minkes-Weiland
Heleen A. Reinders-Messelink
Anne M. Boonstra
Lucas H.V. van der Woude
Rob den Otter

European Journal of physical and rehabilitation medicine, 2022, 58(5), 693

Abstract

Abstract

Background. The Lokomat, one of the most popular robotic exoskeletons, can take the asymmetry in the gait pattern of unilaterally affected patients into account with its opportunity to provide unequal levels of movement support (or 'guidance') to each of the legs. This asymmetrical guidance may be used to (1) selectively unburden limbs with impaired voluntary control and/or (2) to exploit the interlimb couplings for training purposes. However, there is a need to explore and understand these specific device opportunities more broadly before implementing them in training.

Aim. To explore the effects of (a)symmetrical guidance settings on lower limb muscle activity in persons with post stroke hemiparesis, during Lokomat guided gait.

Design. A single group, dependent factorial design

Setting. Rehabilitation center; a single session of Lokomat guided walking.

Population. A group of ten persons with post stroke hemiparesis.

Methods. Participants walked in the Lokomat in eight conditions, consisting of symmetrical and asymmetrical guidance situations, at both 0.28 m/s and 0.56 m/s. During symmetrical conditions, both legs received 30% or 100% guidance, while during asymmetrical conditions one leg received 30% and the other leg 100% guidance. Surface electromyography was bilaterally measured from: Biceps Femoris, Rectus Femoris, Vastus Medialis, Medial Gastrocnemius and Tibialis Anterior. Statistical effects were assessed using Statistical Parametric Mapping.

Results. The provision of assymetrical guidance did not affect the level of lower limb muscle activity. In addition, no effect (except for Vastus Medialis in the affected leg during 1.5 – 2.4% of the gait cycle) of symmetrical guidance on muscle amplitude could be observed.

Conclusions. The results show no evidence that either symmetrical or asymmetrical guidance settings provided by the Lokomat can be used to manipulate activity of lower limb musculature in persons with post stroke hemiparesis.

Clinical rehabilitation impact. This study provides insights for the use of specific opportunities provided by the Lokomat for training purposes post stroke.

Introduction

In western countries, stroke is one of the most common causes of death [1-3] with more prolonged disability than any other condition [2,3]. Stroke survivors frequently experience unilateral motor problems varying from a minor decrease in strength to complete paralysis [2]. Although the legs often recover sufficiently well to allow standing and walking, problems in coordination and balance during gait and ambulation often persist [2]. Therefore, training functional gait ability is an important goal in rehabilitation. During conventional treadmill training, two or three therapists are often required to support the trunk and the affected limb in stepping [4]. As an alternative, Robot-Assisted Gait Trainers (RAGT) have been developed to train the gait ability of persons suffering from stroke by maximizing the training intensity in a task-specific and safe way, while being less strenuous for therapists compared to manual gait training [5,6]. Similar to manual gait training, specific RAGT can take the asymmetry of the gait pattern into account with its opportunity to provide more movement support to one leg compared to the other. However, there is a need to explore and understand these specific opportunities offered by RAGT more broadly before implementing them in training.

The Lokomat is a popular, commercially available treadmill-based robotic exoskeleton that consists of two leg orthoses [7], and offers the possibility to set the level of movement support. This movement support, or ‘guidance’, can be provided based on an impedance control strategy that applies a corrective torque when the trajectory of the leg deviates from the reference trajectory [8,9]. The magnitude of the torque that pushes the leg back to the reference trajectory increases with the amount of guidance provided [9]. Previous research on the neuromuscular control of Lokomat guided gait in able-bodied individuals showed that guidance levels can be used to manipulate the level of muscle activity, and that higher levels of guidance result in lower levels of muscle activity [10,11]. The Lokomat offers the opportunity to set guidance levels for both legs separately or asymmetric. This possibility of setting guidance levels asymmetrically is incorporated into the Lokomat to specifically assist the walker when needed and to take the inherent asymmetries in the gait pattern of certain patients into account.

Asymmetrical movement support opens up opportunities for gait training post stroke. Firstly, an asymmetrical situation can be used to selectively unburden the limb with impaired voluntary control. More support can be provided to the affected leg compared to the unaffected leg to make a physiological gait pattern possible while at the same time allowing more active contributions from the unaffected leg. Secondly, asymmetrical guidance can be used to exploit the interlimb couplings for training purposes. In bipedal locomotion, bilateral leg control is subject to neuromuscular, kinematic, and kinetic couplings [12], so that altered contributions of one leg potentially affect how much the contralateral leg contributes. It can be argued that in situations where asymmetric guidance requires more active contributions from one of the legs, these couplings can facilitate contributions from the contralateral leg. Findings of a previous study in healthy young participants showed that while maximal guidance was offered to the ipsilateral leg, the level of ipsilateral muscle activity could be increased when a higher contribution of the contralateral leg to the gait cycle was demanded by offering less guidance to the contralateral leg [13]. Arguably, by setting guidance asymmetrically during Lokomat training of participants with post stroke hemiparesis, the capacity of the unaffected leg may be utilized to evoke a higher muscle activity in the affected leg. In a training situation, this could be exploited to promote functional muscle activity of the affected leg in a task specific fashion.

Asymmetrical guidance settings can be a promising feature to use in the rehabilitation of hemiparetic patients. Therefore, there is a need to explore and understand these (a)symmetrical guidance settings of the Lokomat. As such, the aim of the present, explorative study was to assess whether (a)symmetrical guidance can influence lower limb muscle activity of the affected and unaffected leg. We will not make claims about the clinical effects of RAGT. Instead, we will explore if and how asymmetrical support can potentially contribute to training that takes the inherent asymmetries in the gait pattern of hemiparetic stroke patients into account. To the best of our knowledge, this study is the first to explore the potential of the specific asymmetrical training opportunities offered by RAGT in participants with post stroke hemiparesis. Since previous studies on Lokomat guided gait showed that effects of guidance on muscle activity depend on treadmill speed [10,11], all effects were mapped on two speed levels separately.

Material and methods

Participants

Ten participants with chronic hemiparetic stroke (7 male, 55.5 ± 12.6 year, 1.73 ± 0.11 meter, 85.6 ± 15.2 kg, 24.4 ± 35.4 months post stroke of which 2 walked with an AFO during the experiment) volunteered in this explorative study. All participants had a first ever unilateral stroke (4 persons had an infarction and 6 had a hemorrhage) that occurred at least three months prior to participation, suffered from a unilateral paresis of the leg (6 participants had a unilateral paresis of the left leg and 4 of the right leg) and had a Functional Ambulation Classification score [14] of 3 (3 participants; *'is capable of walking when a safer environment with supervision or verbal guidance is provided'*), 4 (1 participant; *'can walk independently in and around the house (<200 m) with help of walking aids, on level ground, but requires help when walking > 200 m, on stairs, slopes and uneven surfaces'*) or 5 (6 participants; *'is independently capable of walking on flat and non-flat surfaces, on slopes and is capable of walking the stairs'*). A rehabilitation physician from 'Revalidatie Friesland' (Beetsterzwaag, The Netherlands), where the research took place, recruited the participants based on the information in electronic patient files. A total of 10 patients were included for this explorative study. This number of patients is similar to previous studies concerning the Lokomat [10, 11, 15-18] and therefore seems sufficient for exploring the usefulness of asymmetrical Lokomat settings in rehabilitation after stroke. Participants were screened and excluded when they had severely impaired cognitive functions (Mini Mental State Exam score ≤ 25), severe phatic disorders, communication problems in the Dutch language, visual problems or neglect, or comorbidities that are known to affect the gait pattern. Most of the participants had experience with walking in the Lokomat exoskeleton. The experimental procedures were approved by a Medical Ethical Committee (METc University Medical Center Groningen, the Netherlands; project code: 2017.453) and performed according to the declaration of Helsinki for medical research involving human subjects [19]. All subjects provided written informed consent prior to participation.

Materials

For this study, the Lokomat (LokomatPro version 6.0; Hocoma AG, Volkertswil, Switzerland) located at the rehabilitation center 'Revalidatie Friesland' (Beetsterzwaag, The Netherlands) was used. Besides the adjustable control over guidance levels, the Lokomat offers the opportunity to vary the speed level (0.14–0.89 m/s) and the extent to which body weight support (BWS; 0-100%) is provided.

To assess muscle activity, surface electromyography (EMG) was recorded from five muscles in both legs: Biceps Femoris (BF), Rectus Femoris (RF), Vastus Medialis (VM), Medial Gastrocnemius (MG) and Tibialis Anterior (TA). As previously employed [10,13,15], signals were recorded using Ag/AgCl electrodes (Kendall/Tyco ARBO; Warren, MI, USA) with a 10 mm diameter and a minimum electrode distance of 25 mm. The sensors were used and placed according to the SENIAM guidelines [20]. To ensure good conduction, body hair and dead skin cells were removed and electrode sites were cleaned with alcohol. Custom-made insoles

(Pedag international VIVA, Berlin, Germany), containing four pressure sensors (FSR402, diameter 18 mm, loading 10–1000 g), were used to detect gait events. EMG signals and the pressure sensor data were simultaneously sampled at 2048 Hz. A 32-channel Porti7 portable recording system (Twente Medical Systems, Enschede, The Netherlands) with a common mode rejection of 0.90 dB, a 2 mVpp noise level, and an input impedance of 0.10 GV was used to pre-amplify and A/D convert (22 bits) the signals before storage on a computer for offline processing.

Procedure

Similar to previous studies [13,16], speed levels of 0.28 m/s and 0.56 m/s were provided to the participants. Subjects walked first on the treadmill without the Lokomat exoskeleton during two conditions (0.28 m/s and 0.56 m/s). Then, subjects walked in 16 conditions in the Lokomat, during symmetrical and asymmetrical guidance settings, at both 0.28 m/s and 0.56 m/s. During the experiment, participants walked in the Lokomat with foot straps and without BWS. To ensure that an approximately equal number of steps was measured, conditions at 0.28 m/s lasted 120 seconds and conditions at 0.56 m/s lasted 60 seconds. In symmetrical conditions, both the affected leg and the unaffected leg received 30% guidance (30 affected leg (AL) / 30 unaffected leg (UL)) or 100% guidance (100AL/100UL). During asymmetrical conditions, the level of guidance offered to the unaffected leg was set higher (30AL/100UL) or lower (100AL/30UL) compared to the level of guidance offered to the affected leg. A randomized order of the two treadmill conditions (0.28 m/s and 0.56 m/s) was presented first. Next, the eight unique Lokomat conditions (30AL/30UL, 100AL/100UL, 30AL/100UL and 100AL/30UL, at both 0.28 and 0.56 m/s) were presented in a randomized order (block 1). Finally, the Lokomat conditions were repeated in the reversed order (block 2) to control for learning, order and fatigue effects resulting in a total of 16 Lokomat conditions. An acclimatization period of two minutes was provided to the participants before the start of each trial to get accustomed to the specific Lokomat settings. As body weight support is known to alter the muscle activation pattern in both amplitude and shape [21], no body weight support was provided. Participants walked on their own shoes, without encouragement and instructions and under supervision of a physiotherapist during all conditions.

Data analysis

Offline analysis of the synchronized EMG and pressure sensor data was performed using custom-made software routines in Matlab (Matlab 2016a; MathWorks, Natick, MA). The EMG data were highpass filtered using a 10 Hz fourth order high-pass Butterworth filter to attenuate movement artefacts, full wave rectified, and low-pass filtered using a 20 Hz fourth order Butterworth filter. The EMG signals were time normalized from heelstrike to heelstrike based on the pressure sensor data and amplitude normalized with respect to the maximal amplitude observed during unrestrained treadmill walking at the same speed level, for each muscle and each participant separately. For the analysis, EMG data of the corresponding conditions performed in block 1 and 2 were averaged.

Statistical analysis

Repeated measures ANOVAs were performed using Statistical Parametric Mapping (SPM), which is a relative new approach for analysing continuous biomechanical data [22]. The SPM ANOVA calculates F-values (SPM{F}) separately for each data point in the time domain, meaning no time windows or specific regions of interest need to be defined a priori. Since muscle activity during Lokomat guided gait is observed within time windows that are a-typical compared to unrestrained walking [17], the continuous nature of SPM analysis avoids potential bias through specifically selected time windows.

SPM corrects for familywise error properly by dealing with the spatial correlation present in the continuous EMG signal [23]. Due to the spatial correlation in the signal, the number of independent tests does not equal the number of data points (t=1...100). Therefore, SPM estimates the actual number of independent tests based on temporal gradients and the number of data points to maintain the familywise error rate at the chosen 5% [23]. F-values that exceed the critical threshold are called 'supra-threshold clusters'.

In order to answer the research question, SPM repeated measures ANOVAs with the factors Guidance (30% and 100%) and Symmetry (symmetrical and asymmetrical) were conducted for each leg, muscle and speed level separately. Following the aim, the Guidance by Symmetry interactions were interpreted to assess the effects of asymmetrical guidance. In addition, the main effect of the factor Guidance (30% and 100%) was assessed to map the effects of symmetrical levels of guidance on muscle activity. The direction of the effects was established by comparing the means within the supra-threshold clusters. All SPM analyses were conducted using open-source spm1d code (v.MO.1, www.spm1d.org) in Matlab (R2017b, The MathWorks Inc, Natick, MA).

Data availability

The data reported in this paper are not publicly available but are available from the corresponding author on reasonable request.

Results

The level of lower limb muscle activity is not affected by setting asymmetrical levels of guidance

Fig 1 presents EMG profiles of the affected leg of a single representative participant at 0.28 m/s. As can be observed, muscle activity during Lokomat guided gait is lower compared to treadmill walking. However, the EMG amplitudes during the different guidance settings in the Lokomat do not seem to differ. Consistent with this, the SPM repeated measures ANOVAs showed no significant interaction effects of the factors Guidance and Symmetry (see Fig 2 and Fig 3 for group averaged muscle profiles combined with the SPM{F} values). This indicates that the level of muscle activity in the different asymmetrical guidance conditions was not significantly altered for the both legs.

Symmetrical guidance levels do not have a profound effect on the level of muscle activity

As presented in figure 2 and 3, no supra-threshold clusters for BF, RF, MG and TA for the main effect of Guidance were found. This indicates that a symmetrical change in the applied guidance levels had no effect on the level of muscle activity. One small supra-threshold cluster was found in VM of the affected leg at 0.56 m/s (1.5-2.4% of the gait cycle, P=0.0473; critical threshold = 19.28; see figure 3), indicating that activity decreased during this small period of the gait cycle when guidance increased from 30% to 100%.

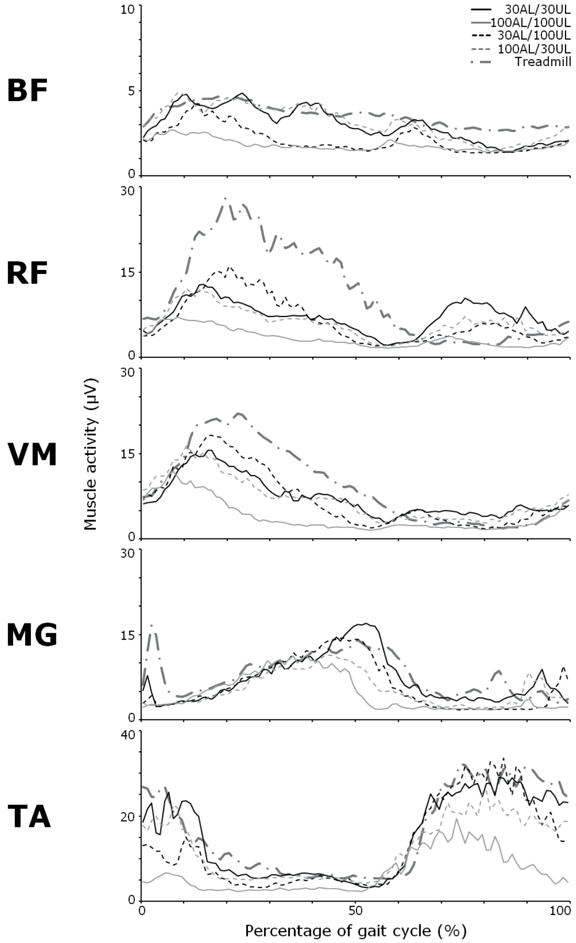

Figure 1. EMG profiles during treadmill walking and Lokomat guided gait. Time normalized EMG profiles for Biceps Femoris (BF), Rectus Femoris (RF), Vastus Medialis (VM), Medial Gastrocnemius (MG) and Tibialis Anterior (TA) for a representative stroke patient at 0.28 m/s.

3

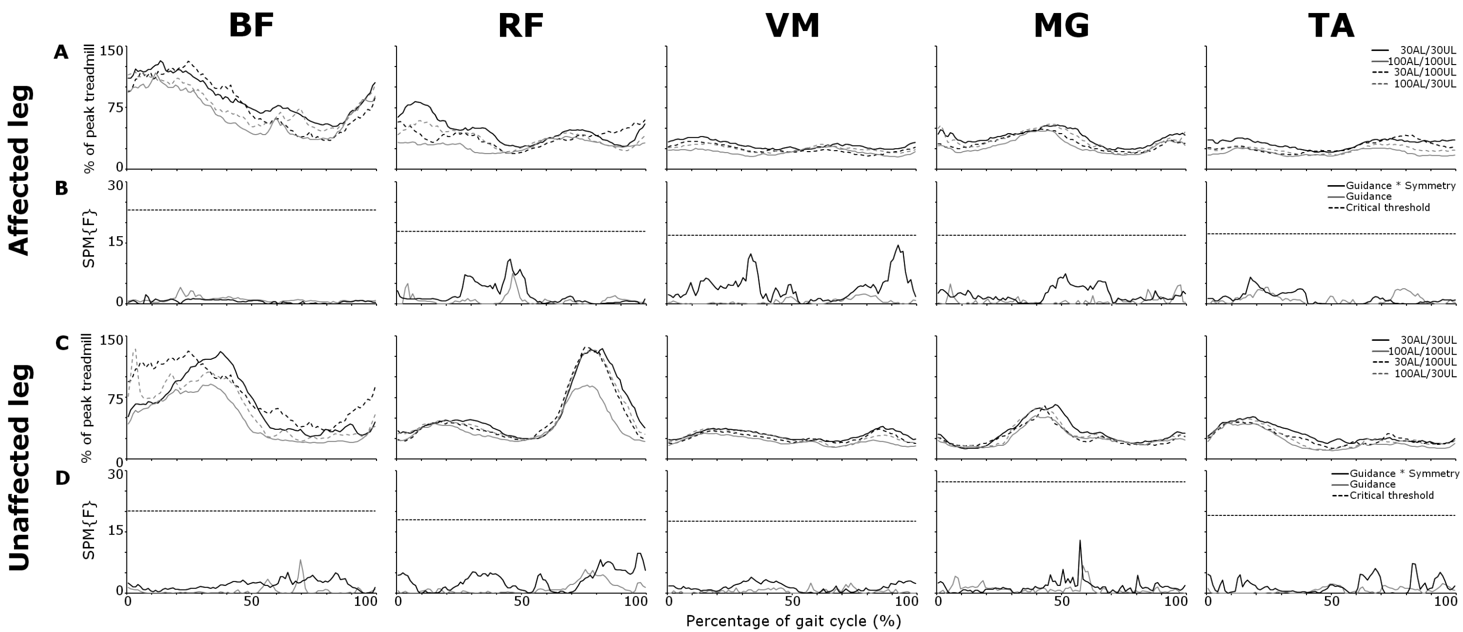

Figure 2. Averaged EMG profiles during Lokomat walking at 0.28 m/s combined with the SPM results (n=10 persons with a chronic hemiparetic stroke). Averaged EMG profiles for Biceps Femoris (BF), Rectus Femoris (RF), Vastus Medialis (VM), Medial Gastrocnemius (MG) and Tibialis Anterior (TA) are time and amplitude normalized for the affected leg (A) and the unaffected leg (C) for both symmetrical (solid lines) and asymmetrical (dashed lines) conditions when the affected leg receives 30% (black lines) or 100 % (grey lines) guidance. EMG amplitude is expressed as a percentage of the peak amplitude during treadmill walking at the same speed level (0.28 m/s). The mean standard deviation for all conditions and all muscles is 14,2 for the affected leg and 15,3 for the unaffected leg. The SPM{F} values are displayed for the Guidance by Symmetry interaction (solid black lines) and the main effect of Guidance (grey lines) for the affected leg (B) and the unaffected leg (D). Threshold F values are indicated by dashed black lines.

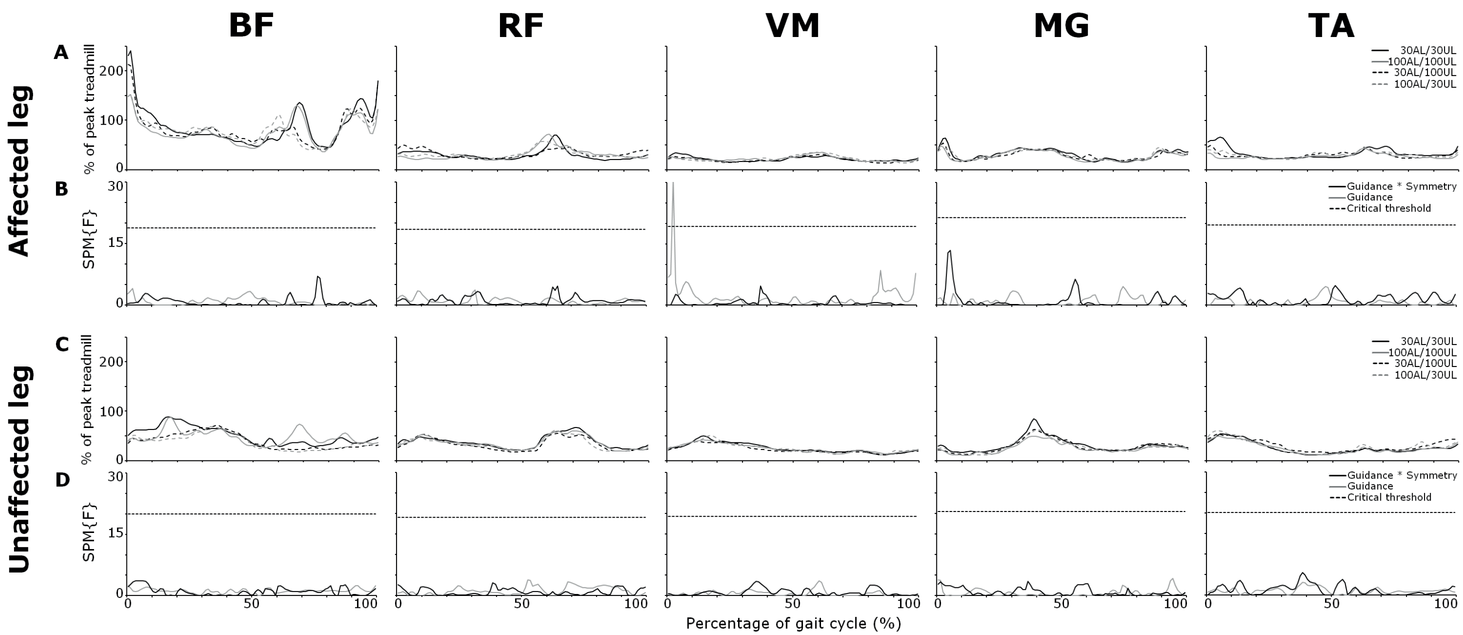

Figure 3. Averaged EMG profiles during Lokomat walking at 0.56 m/s combined with the SPM results (n=10 persons with a chronic hemiparetic stroke). Averaged EMG profiles for Biceps Femoris (BF), Rectus Femoris (RF), Vastus Medialis (VM), Medial Gastrocnemius (MG) and Tibialis Anterior (TA) are time and amplitude normalized for the affected leg (A) and the unaffected leg (C) for both symmetrical (solid lines) and asymmetrical (dashed lines) conditions when the affected leg receives 30% (black lines) or 100 % (grey lines) guidance. EMG amplitude is expressed as a percentage of the peak amplitude during treadmill walking at the same speed level (0.56 m/s). The mean standard deviation for all conditions and all muscles is 14,9 for the affected leg and 16,4 for the unaffected leg. The SPM{F} values are displayed for the Guidance by Symmetry interaction (solid black lines) and the main effect of Guidance (grey lines) for the affected leg (B) and the unaffected leg (D). Threshold F values are indicated by dashed black lines.

Discussion

To the best of our knowledge, this study is the first to explore the effects of asymmetrical guidance on lower limb muscle activity in persons with chronic hemiparetic stroke, during Lokomat guided gait. The possibility of setting guidance levels asymmetrically was incorporated into the Lokomat to take the inherent asymmetries in the gait pattern of certain patient populations into account. If (a) symmetrical guidance can be used to manipulate the level of muscle activity, this can open up opportunities for gait training post stroke. Asymmetrical guidance may than be used to (1) selectively unburden limbs with impaired voluntary control and/or (2) to exploit the interlimb couplings for training purposes. However, as the present results point out, alterations in either symmetrical or asymmetrical guidance settings were generally ineffective in modifying the level of lower limb muscle activity of persons suffering from hemiparetic chronic stroke.

Guidance may not be a reliable feature to use for training as guidance effects seem to be different for healthy able-bodied participants and persons post stroke. Research in healthy participants showed that (a)symmetrical levels of guidance can be used to tune the level of muscle activity [10,11,13]. However, recent research in persons suffering from stroke showed no effect of symmetrical guidance manipulations on the level of muscle activity [23]. In agreement with this, in the present study, no effect of (a)symmetrical guidance on muscle amplitudes in persons suffering from stroke could be observed (except for VM in the affected leg during only 0.9% of the gait cycle). In healthy participants, guidance effects are predominantly present at low speed levels in muscles related to progression of the leg through swing [10,13]. It could be argued that the increase in muscle activity of these muscles with lower levels of guidance are an indirect result of altered leg dynamics due to uncompensated exoskeleton inertia at low speed levels. As the Lokomat does not fully eliminate torques of inertia, friction and gravity [24], more active control may help progress the leg through swing, especially at low levels of guidance and speed. At higher speed levels, effects of guidance in healthy participants are also limited [10,13].

It is not self-evident that the presently used guidance force control mode can be used effectively to influence the level of muscle activity. In the presently used setup, guidance was provided by means of an impedance controller. With maximal guidance, no deviation from the reference trajectory is allowed by the robot [9], meaning that in principle no activity is required for walking. With lower levels of guidance, small deviations from the reference trajectory are allowed [9] creating more movement freedom and allowing more active control from the walker to stimulate autonomous control [25]. However, manipulating the movement freedom with the level of guidance does not always go along with a change in muscular activity. For example, as can be observed in the muscle profiles (Figs 1-3), there is still a phased activity during maximally guided walking, despite the fact that kinematic trajectories in these conditions are fully restricted, and can be produced passively. On the other hand, increasing kinematic freedom, by reducing guidance levels to 30%, did not result in a structural increase in muscular activity. In this context, it is also important to mention that, muscle activity in all Lokomat conditions was remarkably low compared to unrestrained treadmill walking (Figs 1-3). This observation suggests that, irrespective of the level of guidance, the exoskeleton provides sufficient support to produce the required kinematic trajectories with relatively little active muscular involvement.

Other strategies may be more effective to promote a higher level of muscle activity during Lokomat guided gait. In this experiment, participants were not instructed to be more active with lower levels of guidance. As previously shown, children suffering from cerebral palsy increase their active participation when walking in the Lokomat under encouragement [18]. It can be argued that patients, more than healthy subjects, require more encouragement to utilize the additional movement freedom when walking in the Lokomat. The active participation in the Lokomat may also be promoted by using a different control mode. The Lokomat offers a path control mode in Lokomat Pro 6.0 versions [26] that has shown to increase the active participation [24,–26] and kinematic variability [26] during Lokomat guided gait compared to the guidance force control mode. Path control provides the walker temporal and spatial movement freedom inside a virtual tunnel while a supportive force assists in the movement [26]. Since muscle activity is higher when walking with encouragement or in the path control mode compared to the guidance force mode, this may allow for greater adjustments in the level of muscle activity when movement support is (a)symmetrically manipulated.

Generalization of the current results to therapy settings is not self-evident due to several limitations of the study. First, this study explored only the immediate effects of asymmetrical guidance. It is possible that chronic stroke patients need a longer time to adapt to the new situation because of compensation mechanisms already strong developed [27]. Second, the relative low number of participants is a limitation of this explorative study. However, a visual inspection of the individual data, averaged group data (Figs 2-3) and the SPM{F} values (Figs 2-3) indicated no trend of clinically relevant effects of asymmetrical guidance on muscle amplitudes. The visual inspection of the individual data also indicated participants responded divergently on guidance manipulations. Taken together, this means that (a)symmetrical guidance does not seem to be a reliable feature to use in training post stroke for each individual. Nevertheless, the use of the Lokomat for aims like improving blood circulation, lymph circulation and optimization of muscle tone is well established [28]. Lokomat training can thus be effective for patients with specific characteristics for specific training goals.

Conclusions

The present explorative study showed no evidence that (a)symmetrical guidance settings during Lokomat guided gait, evaluated at two different speeds, can be used to manipulate lower limb muscle activity in persons suffering from hemiparetic stroke. No clinically relevant effects in muscular output could be observed as a consequence of either asymmetrical or symmetrical guidance manipulations. Therefore, asymmetrical guidance force should not be used to (1) selectively unburden the limbs with impaired voluntary control and (2) to stimulate interlimb transfers for training purposes.

References

1. Feigin VL, Brainin M, Norrving B, Martins S, Sacco RL, Hacke W, Fisher M, Pandian J, Lindsay P. World Stroke Organization (WSO): global stroke fact sheet 2022. International Journal of Stroke. 2022 Jan;17(1):18-29.
2. Caplan LR, editor. Caplan's stroke. Cambridge University Press; 2016 Sep 8.
3. Feigin VL, Stark BA, Johnson CO, Roth GA, Bisignano C, Abady GG, Abbasifard M, Abbasi-Kangevari M, Abd-Allah F, Abedi V, Abualhasan A. Global, regional, and national burden of stroke and its risk factors, 1990–2019: A systematic analysis for the Global Burden of Disease Study 2019. The Lancet Neurology. 2021 Oct 1;20(10):795-820.
4. Werner C, Von Frankenberg S, Treig T, Konrad M, Hesse S. Treadmill training with partial body weight support and an electromechanical gait trainer for restoration of gait in subacute stroke patients: a randomized crossover study. Stroke. 2002 Dec 1;33(12):2895-901.
5. Kwakkel G, Kollen B, Lindeman E. Understanding the pattern of functional recovery after stroke: facts and theories. Restorative neurology and neuroscience. 2004 Jan 1;22(3-5):281-99.
6. Teasell RW, Foley NC, Bhogal SK, Speechley MR. An evidence-based review of stroke rehabilitation. Topics in stroke Rehabilitation. 2003 Apr 1;10(1):29-58.
7. Calabro RS, Cacciola A, Berte F, Manuli A, Leo A, Bramanti A, Naro A, Milardi D, Bramanti P. Robotic gait rehabilitation and substitution devices in neurological disorders: where are we now?. Neurological Sciences. 2016 Apr 1;37(4):503-14.
8. Hussain S, Xie SQ, Liu G. Robot assisted treadmill training: mechanisms and training strategies. Medical engineering & physics. 2011 Jun 1;33(5):527-33.
9. Riener R, Lünenburger L, Maier IC, Colombo G, Dietz V. Locomotor training in subjects with sensori-motor deficits: an overview of the robotic gait orthosis lokomat. Journal of Healthcare Engineering. 2010;1(2):197-216.
10. van Kammen K, Boonstra AM, van der Woude LH, Reinders-Messelink HA, den Otter R. The combined effects of guidance force, bodyweight support and gait speed on muscle activity during able-bodied walking in the Lokomat. Clinical biomechanics. 2016 Jul 1;36:65-73.
11. Moreno JC, Barroso F, Farina D, Gizzi L, Santos C, Molinari M, Pons JL. Effects of robotic guidance on the coordination of locomotion. Journal of neuroengineering and rehabilitation. 2013 Dec;10(1):79.
12. Arya KN, Pandian S. Interlimb neural coupling: implications for poststroke hemiparesis. Annals of physical and rehabilitation medicine. 2014 Dec 1;57(9-10):696-713.
13. Weiland S, Smit IH, Reinders-Messelink H, van der Woude LH, van Kammen K, den Otter R. The effect of asymmetric movement support on muscle activity during Lokomat guided gait in able-bodied individuals. PloS one. 2018 Jun 4;13(6):e0198473.
14. Holden MK, Gill KM, Magliozzi MR. Gait assessment for neurologically impaired patients: standards for outcome assessment. Physical therapy. 1986 Oct 1;66(10):1530-9.
15. van Kammen K, Boonstra AM, van der Woude LH, Reinders-Messelink HA, den Otter R. Differences in muscle activity and temporal step parameters between Lokomat

guided walking and treadmill walking in post-stroke hemiparetic patients and healthy walkers. Journal of neuroengineering and rehabilitation. 2017 Dec;14(1):32.
16. van Kammen K, Boonstra AM, van der Woude LH, Visscher C, Reinders-Messelink HA, den Otter R. Lokomat guided gait in hemiparetic stroke patients: the effects of training parameters on muscle activity and temporal symmetry. Disability and rehabilitation. 2020 Oct 8;42(21):2977-85.
17. Van Kammen K, Boonstra A, Reinders-Messelink H, den Otter R. The combined effects of body weight support and gait speed on gait related muscle activity: a comparison between walking in the Lokomat exoskeleton and regular treadmill walking. PloS one. 2014 Sep 16;9(9):e107323.
18. Schuler TA, Müller R, Van Hedel HJ. Leg surface electromyography patterns in children with neuro-orthopedic disorders walking on a treadmill unassisted and assisted by a robot with and without encouragement. Journal of neuroengineering and rehabilitation. 2013 Dec;10(1):78.
19. World Medical Association. Ethical principles for medical research involving human subjects. European journal of emergency medicine: official journal of the European Society for Emergency Medicine. 2001 Sep;8(3):221.
20. Freriks B, Hermens H, Disselhorst-Klug C, Rau G. The recommendations for sensor and sensor placement procedures for surface electromyography. In: Hermens H, editors. European recommendations for surface electromyography. Enschede: Roessingh Research and Development 1995:15-53.
21. Cherni Y, Hajizadeh M, Dal Maso F, Turpin NA. Effects of body weight support and guidance force settings on muscle synergy during Lokomat walking. European Journal of Applied Physiology. 2021 Nov;121(11):2967-80.
22. Pataky TC. Generalized n-dimensional biomechanical field analysis using statistical parametric mapping. Journal of biomechanics. 2010 Jul 20;43(10):1976-82.
23. Friston KJ, Ashburner JT, Kiebel SJ, Nichols TE, Penny WD. Statistical paramertric mapping: the analysis of functional brain images. Elsevier; 2007
24. Duschau-Wicke A, von Zitzewitz J, Caprez A, Lunenburger L, Riener R. Path control: a method for patient-cooperative robot-aided gait rehabilitation. IEEE Transactions on Neural Systems and Rehabilitation Engineering. 2009 Oct 6;18(1):38-48.
25. Warken B, Graser JV, Ulrich T, Borggraefe I, Heinen F, Meyer-Heim A, van Hedel HJ, Schroeder AS, Aurich T. Practical recommendations for robot-assisted treadmill therapy (Lokomat) in children with cerebral palsy: indications, goal setting, and clinical implementation within the WHO-ICF framework. Neuropediatrics. 2015 Aug;46(04):248-60.
26. Aurich-Schuler T, Grob F, van Hedel HJ, Labruyère R. Can Lokomat therapy with children and adolescents be improved? An adaptive clinical pilot trial comparing guidance force, path control, and FreeD. Journal of neuroengineering and rehabilitation. 2017 Dec;14(1):76.
27. Mazzoleni S, Battini E, Crecchi R, Posteraro F. Timing of motor recovery in subacute and chronic stroke patients during upper limb robot-assisted rehabilitation. InInternational Conference on NeuroRehabilitation 2018 Oct 16 (pp. 190-194). Springer, Cham.
28. Medvedev IN. Place and possibilities of the robotic system Lokomat in the rehabilitation of patients after ischemic stroke. Biomedical and Pharmacology Journal. 2019 Mar 25;12(1):131-40.

Conflicts of interest: The authors certify that there is no conflict of interest with any financial organization regarding the material discussed in the manuscript.

Funding: This research was funded by a grant from 'Het Revalidatiefonds' (projectnumber: R201705724).

Authors' contributions: SMW, RDO and HRM conceived the experiments. SMW performed the experiments, analysed the data and wrote the manuscript. RDO, HRM, AMB and LVDW were involved in the design of the study, and the drafting and critical revision of the manuscript. All authors read and approved the final version of the manuscript.

Acknowledgements: The authors wish to thank Emyl Smid for his technical support and the physiotherapists Steven Floor, Matthijs Dijkema and Geke van der Heide for their help with the recruitment and data collection.

CHAPTER 4

4

Effects of unilateral swing leg resistance on propulsion and other gait characteristics during treadmill walking in able-bodied individuals

Sylvana Minkes-Weiland
Han Houdijk
Heleen A. Reinders-Messelink
Lucas H.V. van der Woude
Paul P. Hartman
Rob den Otter

Biomechanics, 2025, 5(4),71

Abstract

Background/Objectives: Swing leg resistance may stimulate propulsive force, required for forward progression and leg swing, in post-stroke patients. To assess the potential of swing leg resistance in rehabilitation, more knowledge is needed on how this unilateral manipulation affects gait. Therefore, we explored the bilateral effects of a unilateral swing leg resistance on muscle activity, kinematics, and kinetics of gait in able-bodied individuals.

Methods: Fourteen able-bodied participants (8 female, aged 20.7 ± 0.8 years, BMI 23.5 ± 1.9) walked on an instrumented treadmill at 0.28 m/s, 0.56 m/s, and 0.83 m/s with and without unilateral swing leg resistance provided by a weight (0 kg, 0.5 kg, 1.25 kg, and 2 kg) attached to the leg through a pulley system. Propulsion and braking forces, swing time, step length, transverse ground reaction torques, and muscle activity in the gluteus medius (GM), biceps femoris (BF), rectus femoris (RF), vastus medialis (VM), medial gastrocnemius (MG), and soleus (SOL) were compared between conditions. Statistical analyses were performed using repeated measures ANOVAs, with a significance level of 5%.

Results: Peak propulsive force and propulsive duration increased bilaterally, while peak braking force decreased bilaterally with unilateral swing leg resistance. In addition, the swing time of the perturbed leg increased with swing leg resistance. Muscle activity in the perturbed leg (GM, BF, RF, VM, MG) and the unperturbed leg (GM, BF, VM, MG, SOL) increased. Only in the BF (perturbed leg, late swing) and MG (unperturbed leg, early stance) did the muscle activity decrease with swing leg resistance. No adaptations in step length and transverse ground reaction torques were observed. Specific effects were enhanced by gait speed.

Conclusions: Unilateral swing leg resistance can evoke effects that might stimulate the training of propulsion. A study in post-stroke patients should be conducted to test whether prolonged exposure to unilateral swing leg resistance leads to functional training effects.

Introduction

Walking ability in people after stroke is frequently impaired. One of the factors that is often affected is the ability to generate propulsive forces on the affected side [1]. Propulsion can be defined as the positive anterior component of the ground reaction force (GRF) during the second half of stance [2] and is essential for the forward progression in human gait. A decreased propulsive capacity is often seen in ambulant patients with stroke and is associated with a diminished gait speed, functional balance, long distance walking ability, and self-perceived participation [1,3]. Therefore, improving the propulsive capacity should be an important target in rehabilitation post stroke [4,5].

The provision of restraining forces during walking can be a potential training strategy that is easy, safe, and cheap to stimulate propulsion after stroke [6]. When a resistance force is applied to one leg, more power is required to actively progress the leg through swing, progress the body's center of mass (CoM) forward, and maintain stability. Prolonged exposure to unilateral swing leg resistance can potentially stimulate propulsive capacity in post-stroke patients as it will require the larger bilateral output of several muscles. First, the calf muscles are known to be substantial contributors to the progression of the CoM and the contralateral leg through swing [7,8]. Therefore, a bilateral increase in calf muscle activity is expected with swing leg resistance. Secondly, as previous work showed [9], hip extensors in the unperturbed leg and hip flexors in the perturbed leg play a major role in the forward progression of the trunk and perturbed swing leg, respectively. Increased quadriceps activity in the perturbed leg during swing progresses the leg through swing, while increased hamstring activity in the unperturbed leg contributes to progression of the trunk. These different effects in the perturbed and unperturbed leg may be exploited to target specific gait problems.

Unilateral swing leg resistance can be provided with a pulley force applied to one leg [9,10]. Previous research in participants post stroke and/or healthy participants concerning unilateral swing leg resistance have shown adaptations in the swing phase duration, step length, and hip angles [9,10]. In addition, a study in participants post stroke showed the potential of unilateral swing leg resistance for improving spatial symmetry and gait velocity [10]. However, unilateral swing leg resistance could influence other gait parameters that may not be desirable. For example, the unilateral backward pull might cause larger transverse external moments that need to be at least compensated by the contralateral hip muscles and the upper body to maintain stability. Therefore, there is a need to map the effects of unilateral swing leg resistance on multiple gait characteristics for both the perturbed leg and the unperturbed leg more broadly.

The primary aim of this study was to explore the bilateral effects of unilateral swing leg resistance on (1) anterior–posterior GRF, (2) step parameters, (3) transverse torques, and (4) muscle activity in both the perturbed and unperturbed leg during walking. Previous studies [9,10], primarily focused on the spatial and temporal step parameters, hip angle, and activity in the rectus femoris and hamstrings. In contrast, the present study takes a broader approach by including multiple muscles, the anterior–posterior GRF, and responses in the transverse plane. In this study we included healthy able-bodied individuals to explore basic biomechanical responses, from which we can derive specific manipulations that might be suitable for future gait training interventions in people after stroke. As individuals after stroke often display a decreased gait speed, and propulsion is known to increase with speed [11], it is important to assess how these effects depend on speed. Therefore, the secondary aim was to explore whether the effects of unilateral swing leg resistance were mediated by gait speed.

Methods

Participants

Fourteen healthy able-bodied participants (6 male/8 female, age 20.7 ± 0.8 years, height 1.82 ± 0.84 m, body weight 76.4 ± 9.0 kg, BMI 23.5 ± 1.9) participated in this study. Participants were excluded if they suffered disorders of a physiological, orthopedic, neurological, or sensorimotor nature that could affect the gait pattern. Experimental procedures were approved by the ethics committee of the Department of Human Movement Sciences (UMCG, The Netherlands; 201800977) and performed according to the declaration of Helsinki for medical research involving human subjects [12]. All participants provided written informed consent prior to the procedure.

Experimental protocol

Participants walked on an instrumented treadmill (MGait, Motek, Amsterdam, The Netherlands) under twelve conditions for two minutes each . Data were recorded during the two minute intervals, roughly corresponding to 44–89 gait cycles depending on the applied speed (0.28, 0.56, or 0.83 m/s) and resistance condition (0, 0.5, 1.25, or 2.0 kg). Restraining forces were applied to the ankle during walking using a pulley-based system that was placed behind the treadmill ([13]; see Figure 1). By attaching a weight to a rope that was transferred by the pulley wheels to the subject's right ankle, various resistance forces were applied. In this study, we chose to provide four weight levels (0 kg, 0.5 kg, 1.25 kg, and 2.0 kg) and three speed levels (0.28 m/s, 0.56 m/s, and 0.83 m/s) to account for the range of speeds observed in patients training post stroke [14]. The selected weight levels were based on pilot testing and practical considerations. Our aim was to span a broad range of external resistance, from light to substantially challenging, while ensuring participant safety and feasibility during the treadmill walking. The maximum level of 2.0 kg was close to the upper limit that participants could comfortably tolerate, and the intermediate loads allowed for the systematic assessment of incremental resistance effects. The average added external power generated by the pulley, equal to the product of applied vertical force and gait speed, ranged between 0 and 16.6 W depending on the (speed and weight) conditions. The twelve conditions (four weight levels over the three speed levels) were offered to the participants in a randomized order. Five minute breaks were offered between the conditions as a pragmatic choice to prevent carry-over effects while maintaining a feasible total testing duration. No accommodation period for the specific speed and load settings was provided. During all conditions, participants wore a harness suspended from the ceiling to prevent them from falling onto the belt. No body weight support was provided through this harness. Participants walked in their own shoes and were instructed not to hold the handrails while walking. No further instructions were provided.

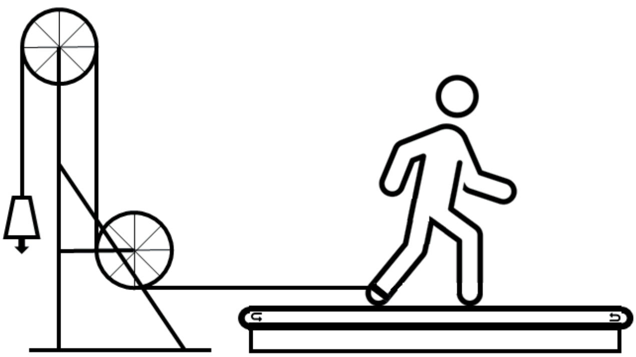

Fig 1. An illustration of the experimental setup. During walking, restraining forces are applied to the ankle using a pulley system positioned behind the treadmill. By attaching weights (0 kg, 0.5 kg, 1.25 kg, and 2 kg in this experiment) to a rope (6.14 m) that runs through the pulleys to the subject's right ankle, different levels of resistance can be introduced.

Data collection

Ground reaction forces (GRF's) and surface electromyography (EMG) were recorded for both the perturbed (right) leg and unperturbed (left) leg. The GRF's were recorded at 1000 Hz using two force plates that were embedded into the treadmill. Surface EMG was recorded for the gluteus medius (GM), biceps femoris (BF), rectus femoris (RF), vastus medialis (VM), medial gastrocnemius (MG), and soleus (SOL) using a Trigno Avanti wireless system (Delsys, Natick, MA, USA; 1925 Hz) with a fixed inter-electrode distance of 10 mm. Electrode locations were determined and prepared based on the SENIAM guidelines [15].

Data analysis

Custom-made software routines were used to analyze the data offline in Matlab (R2015b, MathWorks, Natick, MA, USA). All force plate data were recursively filtered using a 15 Hz second-order low-pass Butterworth filter. Swing time was defined as the time between toe-off and heel strike. Step length, defined as the difference in the anterior–posterior CoP positions at heel strike between legs, was calculated from the force plate data, similar to previous research [16].

The anterior–posterior GRF amplitude was normalized to body weight. Negative values represented braking while positive values represented propulsion [17,18]. The propulsive duration, propulsive peak force, and braking peak force were calculated for each step and each participant. The application of unilateral swing leg resistance could lead to an external rotation moment exerted on the body during a single limb stance. To capture this external rotation moment, data from the force plates was used. The force plates embedded within the treadmill generate moments around three axes based on calibration matrices.

The moment around the longitudinal axis was extracted for each condition and each participant to calculate the mean transverse ground reaction torque during a single stance. This ground reaction torque reflects the sum of the external torque exerted by the pulley and the change in the whole body momentum around the longitudinal axis during a single stance (Newton's second law). Group-averaged force profiles were calculated for the anterior–posterior GRF data and the transverse torques for each condition separately. To establish temporal alignment over the conditions for the statistical analysis, the profiles were time-normalized for stance (60 data points) and swing (40 data points) based on the gait cycle of the perturbed leg.

The EMG data were high-pass filtered using a 10 Hz fourth-order high-pass Butterworth filter to attenuate movement artefacts, full wave rectified, and low-pass filtered using a 20 Hz fourth-order recursive Butterworth filter. The resulting smoothed rectified EMG envelopes of both the perturbed and unperturbed legs were time-normalized for stance (60 data points) and swing (40 data points) of the perturbed leg using the synchronized force plate data, and subsequently averaged over strides, similar to the anterior–posterior GRF. Next, the EMG amplitude was normalized with respect to the maximum amplitude observed during treadmill walking at 0.28 m/s without a restraining force, for each muscle and each participant separately.

Statistical analysis

Repeated measures (RM) ANOVAs with weight (0 kg, 0.5 kg, 1.25 kg, and 2.0 kg) and speed (0.28 m/s, 0.56 m/s, and 0.83 m/s) were conducted to assess the main effect of weight and the interaction of weight and speed for each leg separately. Two statistical approaches were used. Several RM ANOVAs were conducted using SPSS (version 23.0, IBM Corporation, Armonk, NY, USA) to assess the effects on the discrete variables swing time, step length, duration of propulsion, propulsive peak force, braking peak force, and transverse torque. The Benjamini–Hochberg correction was used to correct for multiple testing [19].

For the cycle profiles of anterior–posterior GRF, GM, BF, RF, MG, and SOL, Statistical Parametric Mapping (SPM [20]) was used. Open source spm1d code (v.M0.1), by www.spm1d.org (accessed on 27-12-2022) was used to conduct the SPM RM-ANOVAs in Matlab (R2015b, MathWorks, Natick, MA, USA).

The assumptions for using repeated measures ANOVA were verified by visual inspection of the residual plots to check for normality and homogeneity of variance. Effect sizes (partial eta-squared) were reported for the main analyses. As it was outside the scope of our study, no post hoc tests were conducted.

All tests were conducted separately for both the perturbed and unperturbed leg with an alpha of 0.05. The main effects of weight were interpreted according to our primary research aim 'to explore the immediate bilateral effects of unilateral swing leg resistance on (1) anterior-posterior GRF, (2) step parameters, (3) transverse torques, and (4) muscle activity for both the perturbed and unperturbed leg during walking'. To answer the secondary research question 'to explore whether functional effects of unilateral swing leg resistance were mediated by gait speed', the interactive effects of weight with speed were interpreted. In case of significant effects or supra-threshold clusters, the direction of the effects was determined by comparing the means within the clusters.

Results

An overview of all SPM test results, including the main effects of speed, is presented in Tables A1 and A2 in Appendix A. The main effects of speed are illustrated in Figures A1–A3 in Appendix A; however, similar to the post hoc tests, they are not dis-cussed in the manuscript as this was beyond the scope of the current study.

Anterior-posterior GRF

The braking peak force decreased in both the perturbed and unperturbed legs with unilateral swing leg resistance, while the propulsive peak force bilaterally increased with unilateral swing leg resistance (see Table 1 and Figures 2 and 3). The significant weight–speed interactions indicated that these effects on the braking peak force in the perturbed leg were diminished with speed, while the effects on the braking peak force in the unperturbed leg and the propulsive peak force in both legs were enhanced at higher gait speeds (Figure 2).

The duration of the propulsive phase increased in both legs with unilateral swing leg resistance, as evidenced by a significant main effect of weight (Figure 2). A significant weight–speed interaction for the perturbed leg indicated that the effect of unilateral swing leg resistance on propulsive duration in the perturbed leg reduced with higher speeds.

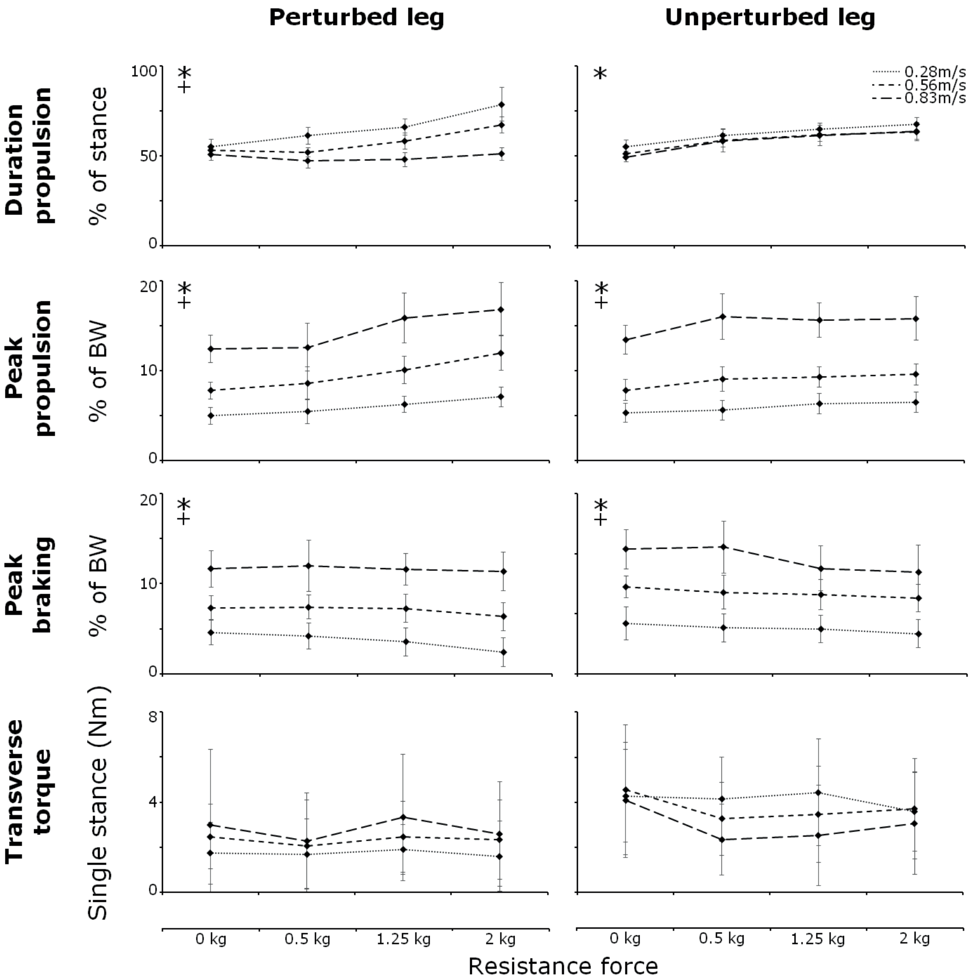

Fig 2. The duration of propulsion, peak propulsive force, peak braking force and transverse torques during walking with and without swing leg resistance. Group-averaged means and standard deviations are presented during walking in all conditions consisting of four resistance forces (provided by attaching a weight of 0 kg, 0.5 kg, 1.25 kg and 2 kg to the pulley-based system) and three speed levels (0.28 m/s, 0.56 m/s and 0.83 m/s). * indicates a significant effect of the factor Weight and $^{+}$ indicates a significant interaction effect of the factor Weight with the factor Speed.

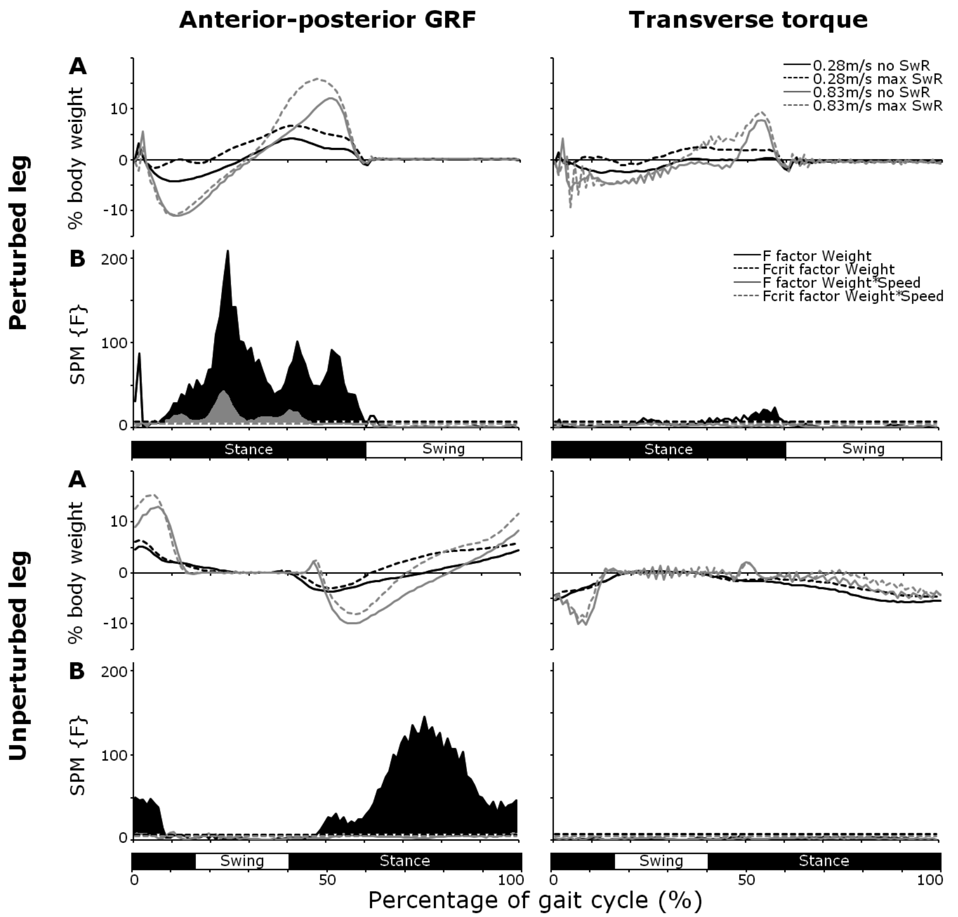

Fig 3. Profiles of the anterior-posterior GRF and transverse torque during walking with and without swing leg resistance together with the results of the SPM analysis. Group-averaged profiles of for walking with (dashed lines) and without (solid lines) swing leg resistance (SwR) at 0.28 m/s (black lines) and at 0.83 m/s (grey lines) (A) combined with the SPM F-values of the factor Weight (black lines) and the Weight by Symmetry interaction (grey lines) (B). Critical threshold F values (Fcrit) are indicated in part B of the figure by dashed lines and significant supra-threshold clusters are indicated by a black fill (for the factor Weight) or a grey fill (for the Weight by Symmetry interaction).

Table 1: Effects of swing leg resistance in the perturbed and unperturbed leg

	Perturbed leg						Unperturbed leg					
	Weight		Speed		Weight*Speed		Weight		Speed		Weight*Speed	
	F(3,39)	η_p^2	F(2,26)	η_p^2	F(6,78)	η_p^2	F(3,39)	η_p^2	F(2,26)	η_p^2	F(6,78)	η_p^2
Peak braking	**14.3****	0.5	211.4**	0.9	**2.7***	0.2	**18.9****	0.6	179.5**	0.9	**4.4****	0.3
Peak propulsion	**52.1****	0.8	310.3**	1.0	**3.3***	0.2	**46.6****	0.8	305.1**	1.0	**4.7****	0.3
Duration propulsion	**90.8****	0.9	82.6**	0.9	**33.0****	0.7	**128.9****	0.9	22.9**	0.6	1.0	0.1
Swing time	**40.1****	0.8	38.8**	0.7	3.8	0.2	1.7	0.1	39.3**	0.7	1.6	0.2
Step length	3.4	0.2	220.5**	0.9	1.6	0.1	0.8	0.1	228.4**	0.9	1.7	0.1
Transverse ground reaction torque	0.8	0.1	3.3	0.2	0.3	0.0	3.6	0.2	6.9*	0.3	2.5	0.2

Note. This table displays the univariate F values of the repeated measures ANOVAs for both legs together with the partial eta-squared effect sizes (η_p^2). Significant results that are discussed in the text are presented in bold (*=p<0.05, **=p<0.001).

Step parameters

As presented in Table 1 and Figure 4, a significant main effect of the factor weight was that the swing time of the perturbed leg increased with leg resistance, as indicated by a significant main effect of weight (Figure 4). No effects on step length in both legs, swing time of the unperturbed leg, and/or significant weight–speed interactions on these parameters were found.

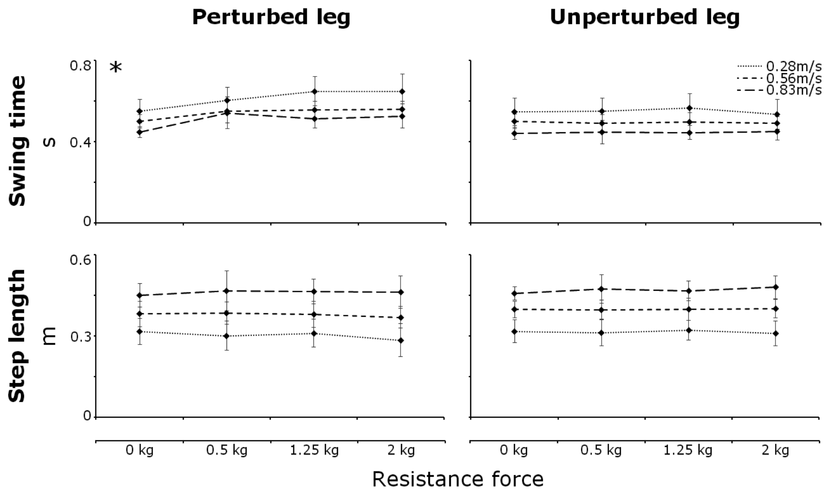

Fig 4. Swing time and step length during walking with and without swing leg resistance. Group-averaged means and standard deviations are presented during walking in all conditions consisting of four resistance forces (provided by attaching 0 kg, 0.5 kg, 1.25 kg and 2 kg to the pulley-based system) and three speed levels (0.28 m/s, 0.56 m/s and 0.83 m/s). * indicates a significant effect of the factor Weight and $^{+}$ indicates a significant interaction effect of the factor Weight with the factor Speed.

Transverse torques

The RM ANOVAs showed no significant effects of unilateral swing leg resistance on the mean transverse torques during single stance (see figure 2 for means and standard deviations). As presented in figure 3, the SPM analysis showed only one small supra-threshold cluster for the factor Weight in the perturbed leg during late stance indicating a small increase of transverse torque with swing leg resistance.

Muscle activity

Group-averaged muscle activity profiles for both legs are presented in Figure 5 (GM, BF, and RF) and Figure 6 (VM, MG, and SOL). As can be observed, several supra-threshold clusters were found for the main effect of the factor weight, indicating that the muscle activity in selected muscles of the perturbed and unperturbed leg were affected by unilateral swing leg resistance. Muscle activation amplitudes increased within most of the supra-threshold clusters when swing leg resistance was provided. Only in BF (perturbed leg, late swing) and MG (unperturbed leg, early stance) did the muscle amplitudes decrease with unilateral swing leg resistance.

As presented in Figures 5 and 6, several small significant supra-threshold clusters for the weight–speed interaction were found, indicating that the effect of the resistance force on the muscle activation amplitude was different over speeds. As can be observed in Figure 5, the effects of weight on BF (perturbed leg, late swing), RF (perturbed leg, mid-swing), VM (perturbed leg, late stance), and BF (unperturbed leg, mid-stance) were larger at higher speeds while the effect on RF (perturbed leg, late swing) was smaller at higher speeds.

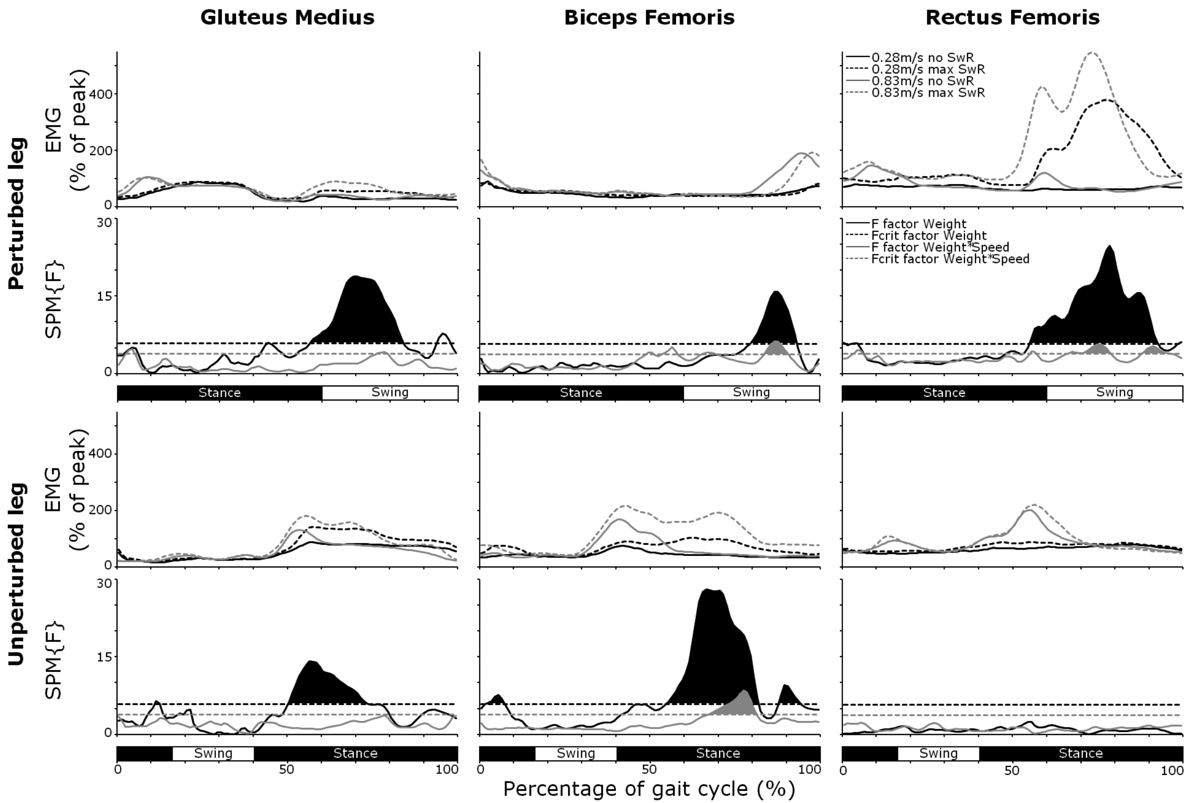

Fig 5. Profiles of Gluteus Medius, Biceps Femoris and Rectus Femoris during walking with and without swing leg resistance together with the results of the SPM analysis. Group-averaged EMG profiles for walking with (dashed lines) and without (solid lines) swing leg resistance (SwR) at 0.28 m/s (black lines) and at 0.83 m/s (grey lines) combined with the SPM F-values of the factor Weight (black lines) and the Weight by Symmetry interaction (grey lines). Critical threshold F values (Fcrit) are indicated by dashed lines and significant supra-threshold clusters are indicated by a black fill (for the factor Weight) or a grey fill (for the Weight by Symmetry interaction).

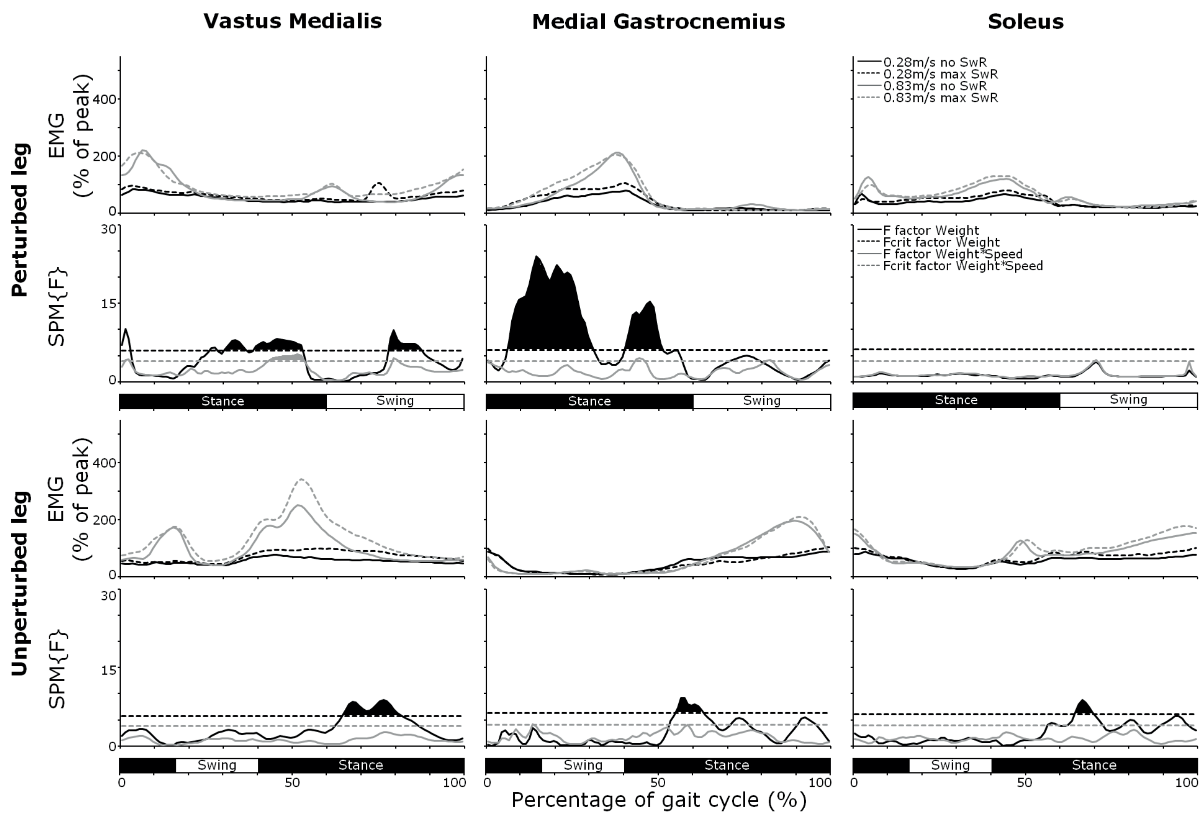

Fig 6. Profiles of Vastus Medialis, Gastrocnemius Medialis and Soleus during walking with and without swing leg resistance together with the results of the SPM analysis. Group-averaged EMG profiles for walking with (dashed lines) and without (solid lines) swing leg resistance (SwR) at 0.28 m/s (black lines) and at 0.83 m/s (grey lines) combined with the SPM F-values of the factor Weight (black lines) and the Weight by Symmetry interaction (grey lines). Critical threshold F values (Fcrit) are indicated by dashed lines and significant supra-threshold clusters are indicated by a black fill (for the factor Weight) or a grey fill (for the Weight by Symmetry interaction).

4

Discussion

The primary aim of this study was to explore the immediate bilateral effects of unilateral swing leg resistance on (1) anterior–posterior GRF, (2) step parameters, (3) transverse torques, and (4) muscle activity. The secondary aim was to explore whether these effects were mediated by gait speed.

Similar to previous research [9], the swing time of the perturbed leg increased with swing leg resistance. This suggests that the perturbed passive swing movement requires active control. While the temporal asymmetry increased, our results suggest that spatial symmetry is maintained in able-bodied participants when walking with unilateral swing leg resistance as no effects on step length could be observed. This is in contrast to previous research showing that step length decreased in the perturbed leg and increased in the unperturbed leg with unilateral swing leg resistance [9]. One possible explanation is the walking speed: a previous study [9] tested at 1.34 m/s, whereas our participants walked at a maximum of 0.83 m/s, which may reduce the spatial effects and account for the lack of step length changes in our study.

We hypothesized that propulsion would increase during unilateral swing leg resistance to enable forward progression. The present results show that such effects were observed in both the perturbed and unperturbed leg; propulsion increased, and braking decreased bilaterally. The observed changes in muscle activation likely accommodated the increased demand for progression of the trunk and the perturbed swing leg. In the perturbed leg, a small increase in VM activity and a substantial increase in RF activity, during late stance and throughout swing, respectively, assists to progress the perturbed swing leg forward. In addition, the increase in MG activity in the perturbed leg during late stance contributes to the ankle push-off that assists in progression of the perturbed leg through swing [21]. In the unperturbed leg, the BF seems primarily responsible for the increase in propulsion. The substantial increase in BF activity in the unperturbed leg during stance extends the hip and drives the trunk forward. By extending the hip, the proportion of the GRF that is distributed anteriorly (the propulsive force) can be increased [2]. The decrease in BF activity in the perturbed leg during late swing is a direct effect of the resistance force that diminishes the need for active braking. Furthermore, a subtle increase in unperturbed SOL and MG activity could be observed primarily during single stance in the unperturbed leg, which ensures vertical support and may assist in the forward progression of the trunk [21].

The current results showed no extreme transverse ground reaction torques, suggesting that the body keeps deviations in the transverse plane to a minimum. This might be mitigated by the reciprocal activation of the GM in both legs that we observed. These bilateral muscles can work together to maintain stability during the perturbed swing phase to compensate for the external torsion (see Figure 5). In addition, adaptations in the upper body movement (transverse plane trunk rotations and arm swing) [22] and in step width (by decreasing step width, the transverse plane moment arm of the impeding force relative to the stance foot can be minimized) might be involved in regulating the transverse torques and hence in maintaining balance and stability as well. However, we did not collect the data to investigate these additional compensatory mechanisms.

Taken together, these findings in able-bodied individuals provide more knowledge on how unilateral swing leg resistance affects the ‘normal’ gait pattern that is not affected by certain health conditions and demonstrate that unilateral swing leg resistance can evoke effects that might stimulate the training of propulsion. An important question for clinical translation is whether swing leg resistance should be applied to the affected leg, the unaffected leg, or both. A follow-up study should be conducted to determine the effects of unilateral swing leg resistance in patients post stroke. If similar effects can be established in patients post stroke, it may be relevant to apply unilateral swing leg resistance to the affected or unaffected leg, depending on the specific patient needs. Our results show several effects in the perturbed leg that may be relevant for the training of the affected leg post stroke. First, as the perturbed leg showed a subtle increase in MG activity in late stance while walking with the resistance, unilateral swing leg resistance offered to the affected leg may stimulate push-off by the calf muscles in the affected leg in a task-specific context. This may be relevant post stroke as muscle weakness in the calf muscle is one of the main factors responsible for the decline in propulsive capacity [3]. Second, as the perturbed leg showed a substantial increase in RF activity during swing while walking with the resistance, unilateral swing leg resistance can potentially stimulate the muscle activity that can enable the forward progression of the swing leg. This may be relevant in training as swing leg progression is often limited post stroke [23]. Third, although this was not tested statistically, peak propulsion seems to be stimulated to a greater extent in the perturbed leg (Figure 3), suggesting that propulsion can be further increased when unilateral swing leg resistance is offered to the affected leg. On the other hand, the unperturbed leg showed a substantial increase in BF activity during stance. For specific patients who may not be able to restitute functional gait characteristics in the affected leg post stroke (i.e., calf muscle activity for push-off), it may be of functional importance to apply unilateral swing leg resistance to the unaffected leg so that the affected leg BF activity is stimulated. The present results urge further research into the effects of unilateral swing leg resistance in post-stroke patients to determine the potential of swing leg resistance in training situations.

Several effects of unilateral swing leg resistance were mediated by gait speed. In training settings, it may be relevant that the stimulation of peak propulsion in both legs can be further increased with gait speed.

The direct effects established here demonstrate how unilateral swing leg resistance affects the neuromechanics of normal gait. From this, potentially useful stimuli for training gait function after stroke could be derived. The clinical implications, however, are not straightforward. First, the increase in MG activity (early stance) in the perturbed leg, where generally no activity can be observed during normal walking, may be irrelevant to stimulate. Second, our data show the earlier onset of the propulsive phase during single stance (see Figure 4) that may be undesirable. The compensatory shift of propulsion from double support to single stance involves more work from the hip muscles (see the large increase in unperturbed BF activity with unilateral swing leg resistance in Figure 5) which is less energy efficient [24]. As individuals after stroke already experience an increased energy expenditure during walking [25], it may not be beneficial to stimulate this propulsive shift, since further elevating the energy cost could limit the feasibility and sustainability of such an intervention in rehabilitation practice. On the other hand, in certain training contexts, improving propulsion may outweigh the drawbacks of the higher energy demands, particularly if short bouts of targeted practice can induce beneficial neuromuscular adaptations. Further research is therefore needed to clarify how the balance between propulsion gains and energy cost influences the feasibility and effectiveness of swing leg resistance training in stroke rehabilitation.

The current study provides a broad view of the bilateral effects of unilateral swing leg resistance during treadmill walking with the selection of muscles, kinetic, and kinematic variables recorded. One limitation, however, is that we did not include measures of the step width, upper body kinematics, and joint angles. These elements may contribute to the understanding of compensatory strategies (e.g., altered joint angles or forward trunk lean) when walking under unilateral loading. Future studies should therefore incorporate upper body dynamics to provide a more complete understanding of whole-body adaptations to swing leg resistance. Additionally, this study was designed to capture the immediate effects of resistance during relatively short two minute bouts of walking under uncommon gait conditions. Since training likely involves longer-term motor learning and adaptation processes that exceed two minutes of practice, future studies should investigate both the prolonged exposure and retention effects to better understand the potential of unilateral resistance training on human gait in rehabilitation.

Conclusions

In conclusion, the present study demonstrates that unilateral swing leg resistance can evoke effects that might stimulate the training of propulsion. Within the current protocol, the applied resistance levels (0, 0.5, 1.25, or 2.0 kg) and walking speeds (0.28 m/s, 0.56 m/s, and 0.83 m/s) were well tolerated and did not induce signs of fatigue in our healthy young participants. A study in the target population should be conducted to investigate whether prolonged exposure to unilateral swing leg resistance leads to functional training effects.

References

1. Awad, L.N.; Reisman, D.S.; Kesar, T.M.; Binder-Macleod, S.A. Targeting paretic propulsion to improve poststroke walking function: A preliminary study. Arch. Phys. Med. Rehabil. 2014, 95, 840–848.
2. Hsiao, H.; Knarr, B.A.; Pohlig, R.T.; Higginson, J.S.; Binder-Macleod, S.A. Mechanisms used to increase peak propulsive force following 12-weeks of gait training in individuals poststroke. J. Biomech. 2016, 49, 388–395.
3. Nadeau, S.; Gravel, D.; Arsenault, A.B.; Bourbonnais, D. Plantarflexor weakness as a limiting factor of gait speed in stroke subjects and the compensating role of hip flexors. Clin. Biomech. 1999, 14, 125–135.
4. Mauritz, K.H. Gait training in hemiplegia. Eur. J. Neurol. 2002, 9, 23–29.
5. Roelker, S.A.; Bowden, M.G.; Kautz, S.A.; Neptune, R.R. Paretic propulsion as a measure of walking performance and functional motor recovery post-stroke: A review. Gait Posture 2019, 68, 6–14.
6. Awad, L.N.; Lewek, M.D.; Kesar, T.M.; Franz, J.R.; Bowden, M.G. These legs were made for propulsion: Advancing the diagnosis and treatment of post-stroke propulsion deficits. J. Neuroeng. Rehabil. 2020, 17, 139 .
7. Hof, A.L.; Nauta, J.; van der Knaap, E.R.; Schallig, M.A.; Struwe, D.P. Calf muscle work and segment energy changes in human treadmill walking. J. Electromyogr. Kinesiol. 1992, 2, 203–216.
8. Meinders, M.; Gitter, A.; Czerniecki, J.M. The role of ankle plantar flexor muscle work during walking. Scand. J. Rehabil. Med. 1998, 30, 39–46.
9. Savin, D.N.; Tseng, S.C.; Morton, S.M. Bilateral adaptation during locomotion following a unilaterally applied resistance to swing in nondisabled adults. J. Neurophysiol. 2010, 104, 3600–3611.
10. Savin, D.N.; Morton, S.M.; Whitall, J. Generalization of improved step length symmetry from treadmill to overground walking in persons with stroke and hemiparesis. Clin. Neurophysiol. 2014, 125, 1012–1020.
11. Jonkers, I.; Delp, S.; Patten, C. Capacity to increase walking speed is limited by impaired hip and ankle power generation in lower functioning persons post-stroke. Gait Posture 2009, 29, 129–137.
12. World Medical Association. World Medical Association Declaration of Helsinki. Ethical principles for medical research involving human subjects. Bull. World Health Organ. 2001, 79, 373.
13. Veeger, D.H.E.J.; Van der Woude, L.H.; Rozendal, R.H. The effect of rear wheel camber in manual wheelchair propulsion. J. Rehabil. Res. Dev. 1989, 26, 37–46.
14. Blokland, I.; Gravesteijn, A.; Busse, M.; Groot, F.; van Bennekom, C.; van Dieen, J.; de Koning , J.; Houdijk, H. The relationship between relative aerobic load, energy cost, and speed of walking in individuals post-stroke. Gait Posture 2021, 89, 193–199.
15. Hermens, H.J.; Freriks, B.; Merletti, R.; Stegeman, D.; Blok, J.; Rau, G.; Disselhorst-Klug, C. European recommendations for surface electromyography. Roessingh Res. Dev. 1999, 8, 13–54.
16. Swart, S.B.; den Otter, R.; Lamoth, C.J. Anticipatory control of human gait following simulated slip exposure. Sci. Rep. 2020, 10, 9599.
17. Donno, L.; Monoli, C.; Frigo, C.A.; Galli, M. Forward and backward walking: Multifactorial characterization of gait parameters. Sensors 2023, 23, 4671.

18. Turns, L.J.; Neptune, R.R.; Kautz, S.A. Relationships between muscle activity and anteroposterior ground reaction forces in hemiparetic walking. Arch. Phys. Med. Rehabil. 2007, 88, 1127–1135.
19. Benjamini, Y.; Hochberg, Y. Controlling the false discovery rate: A practical and powerful approach to multiple testing. J. R. Stat. Soc. Ser. B Methodol. 1995, 57, 289–300.
20. Pataky, T.C. Generalized n-dimensional biomechanical field analysis using statistical parametric mapping. J. Biomech. 2010, 43, 1976–1982.
21. Neptune, R.R.; Kautz, S.A.; Zajac, F.E. Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. J. Biomech. 2001, 34, 1387–1398.
22. De Graaf, M.L.; Hubert, J.; Houdijk, H.; Bruijn, S.M. Influence of arm swing on cost of transport during walking. Biol. Open 2019, 8, bio039263.
23. Dean, J.C.; Bowden, M.G.; Kelly, A.L.; Kautz, S.A. Altered post-stroke propulsion is related to paretic swing phase kinematics. Clin. Biomech. 2020, 72, 24–30.
24. Oh, K.; Baek, J.; Park, S. Gait strategy changes with acceleration to accommodate the biomechanical constraint on push-off propulsion. J. Biomech. 2012, 45, 2920–2926.
25. Awad, L.N.; Palmer, J.A.; Pohlig, R.T.; Binder-Macleod, S.A.; Reisman, D.S. Walking speed and step length asymmetry modify the energy cost of walking after stroke. Neurorehabilit. Neural Repair 2015, 29, 416–423.

Funding: This research received no external funding.

Institutional Review Board Statement: The study was conducted in accordance with the Decla-ration of Helsinki and approved by the Ethics Committee of the Department of Human Movement Sciences (UMCG, The Netherlands; protocol code 201800977, date of approval 14-02-2019).

Informed Consent Statement: Informed consent was obtained from all subjects involved in the study.

Data Availability Statement: The original contributions presented in this study are included in the article. Further inquiries can be directed to the corresponding author(s).

Acknowledgments: The authors wish to thank Lars Dijk and Eline Zaal for their help with the recruitment and data collection of participants. In addition, the authors wish to thank the technical support of the department of Human Movement Sciences of the University Medical Center Groningen, especially Emyl Smid and Anniek Heerschop, for their technical help and advice.

Conflicts of Interest: The authors declare no conflicts of interest.

Supplementary material

A complete overview of the SPM (Statistical Parametric Mapping) test results are presented in Table A1 (for anterior–posterior GRF and the transverse torque) and Table A2 (for muscle activity).

Figure A1 (for anterior–posterior GRF and transverse torque), Figure A2 (for gluteus medius, biceps femoris, and rectus femoris) and Figure A3 (for vastus medialis, medial gastrocnemius, and soleus) present the main effects of speed of the different SPM analyses.

Table A1 Overview of the SPM results for anterior-posterior GRF and the transverse torque

	Perturbed leg				Unperturbed leg			
	Fcrit	Weight (3,39)	Speed(2,26)	Weight* Speed(6,78)	Fcrit	Weight (3,39)	Speed(2,26)	Weight* Speed(6,78)
A-P GRF	Weight: 6.8 Speed: 9.4 Weight*Speed: 4.3	0-1.9*	0.2-2.2*	1.4-2.3*	Weight: 6.4 Speed: 8.8 Weight*Speed: 4.1	0.0-7.7**	0.0-16.2**	0.0-5.1*
		4.7-5.4*	3.3-35.0**	3.6-46.3**		47.1-99.0**	36.7-37.6*	7.6-10.9*
		6.3-59.5**	38.5-56.0**	49.5-54.8**			43.0-48.5**	16.7-21.6*
		60.4-62.8*	57.9-58.1*	59.3-61.5*			49.6-77.5**	51.6-52.8*
			60.4-62.0*				83.8-99.0**	53.0-62.7**
								80.7-87.5**
								88.9-90.2*
								91.6-92.8*
								94.2-99.0*
Transverse torque	Weight: 7.0 Speed: 9.7 Weight*Speed: 4.4	0.7-1.3*	0.3-1.5*	1.6-3.0*	Weight: 6.5 Speed: 8.9 Weight*Speed: 4.2		3.3-10.6**	9.8-10.2*
		22.1-27.2**	3.9-4.0*	23.2-24.5*			13.3-14.3*	
		28.8-29.1*	5.6-6.4*	25.4-26.6*			40.3-41.8*	
		37.7-38.4*	7.9-8.1*	28.0-28.0*			42.3-43.6*	
		39.4-40.8*	10.0-34.8**	59.7-60.4*			48.3-51.6**	
		41.2-48.0**	36.0-36.0*				59.8-60.1*	
		48.0-59.4**	47.9-56.1**				61.7-62.3*	
		60.9-61.0*	60.1-62.4**				67.2-95.3**	

Note. This table displays the supra-threshold clusters with its p-value (*=p<0.05, **=p<0.001) together with the critical threshold values (Fcrit) for both main effects and their interaction effect.

Table A2 Overview of the SPM results for muscle activity

	Perturbed leg				Unperturbed leg			
	Fcrit	Weight (3,39)	Speed(2,26)	Weight*Speed(6,78)	Fcrit	Weight (3,39)	Speed(2,26)	Weight*Speed(6,78)
GM	Weight: 5.8 Speed: 7.8 Weight*Speed: 3.8	43.8-44.3*	0.7-12.9**	2.1-4.9*	Weight: 5.9 Speed: 8.0 Weight*Speed: 3.9	10.6-12.0*	0-2.7*	
		56.1-83.8**	65.6-78.4**	75.7-78.9*		49.7-73.1**	12.9-20.2*	
		93.4-97.5*					31.0-39.6*	
							47.1-57.6**	
							93.5-99.0*	
BF	Weight: 5.7 Speed: 7.7 Weight*Speed: 3.8	78.8-92.7**	0.0-1.5*	48.4-50.6*	Weight: 5.9 Speed: 7.9 Weight*Speed: 3.8	1.2-6.8*	31.2-54.9**	66.7-80.8**
			2.8-7.8*	53.2-57.6*		54.2-81.5**	59.3-82.6**	
			10.0-14.3*	64.7-68.7*		87.2-93.9*		
			37.6-52.0**	83.0-90.1*				
			65.0-99.0**					
RF	Weight: 5.8 Speed: 7.8 Weight*Speed: 3.8	0.0-0.5*	0.0-19.3**	2.7-5.7*	Weight: 5.7 Speed: 7.7 Weight*Speed: 3.8		8.2-14.0*	
		2.2-4.1*	47.1-50.4*	54.5-56.5*			47.4-50.7*	
		54.4-92.2**	55.7-79.3**	65.3-78.4**			58.0-61.8*	
		98.0-99.0*	97.8-99.0*	87.4-96.1*				
VM	Weight: 5.9 Speed: 8.0 Weight*Speed: 3.9	0.0-2.5*	0.0-2.6*	0.8-2.3*	Weight: 5.7 Speed: 7.7 Weight*Speed: 3.8	63.9-81.3**	14.9-15.2*	
		25.5-27.8*	12.6-17.7*	42.4-53.2**			38.3-75.1**	
		29.7-36.2*	40.0-58.5**	78.6-80.4*				
		38.1-52.5**	59.3-63.1*					
		77.5-87.2**	78.6-80.4*					

4

	Perturbed leg				Unperturbed leg			
			84.6-99.0**					
MG	Weight: 6.0 Speed: 8.2 Weight*Speed: 3.9	5.5-30.7**	0.0-5.2**	42.6-45.4*	Weight: 6.3 Speed: 8.6 Weight*Speed: 4.0	54.3-62.7**	0.0-9.8**	12.9-13.2*
		39.6-50.9**	17.8-44.0**	80.6-82.4*			46.0-57.5**	
			72.0-80.9**				65.4-99.0**	
			98.5-99.0*					
SOL	Weight: 6.2 Speed: 8.4 Weight*Speed: 4.0		24.4-51.8**	69.6-70.3*	Weight: 6.1 Speed: 8.2 Weight*Speed: 3.9	63.8-69.4*	0.0-4.7*	
			68.0-70.0*				44.3-48.1*	
			96.5-98.4*				54.6-55.4*	
							60.5-99.0**	

Note. This table displays the supra-threshold clusters with its p-value (*=p<0.05, **=p<0.001) together with the critical threshold values (Fcrit) for both main effects and their interaction effect.

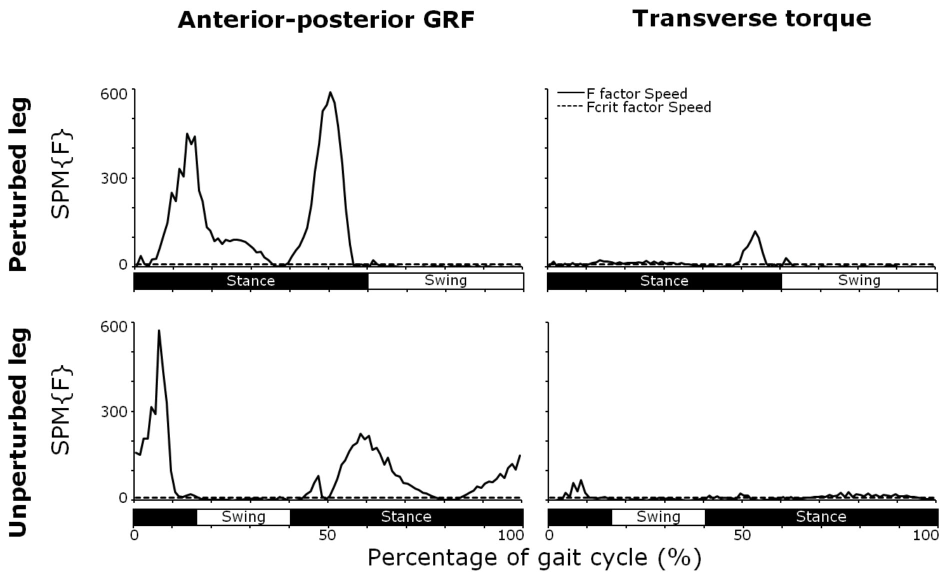

Figure A1. Main effects of Speed for anterior-posterior GRF and transverse torque. SPM F-values are presented for the main effect of Speed (black lines) together with the critical threshold F values (Fcrit; dashed lines). Data points that exceed the Fcrit represent the significant supra-threshold clusters.

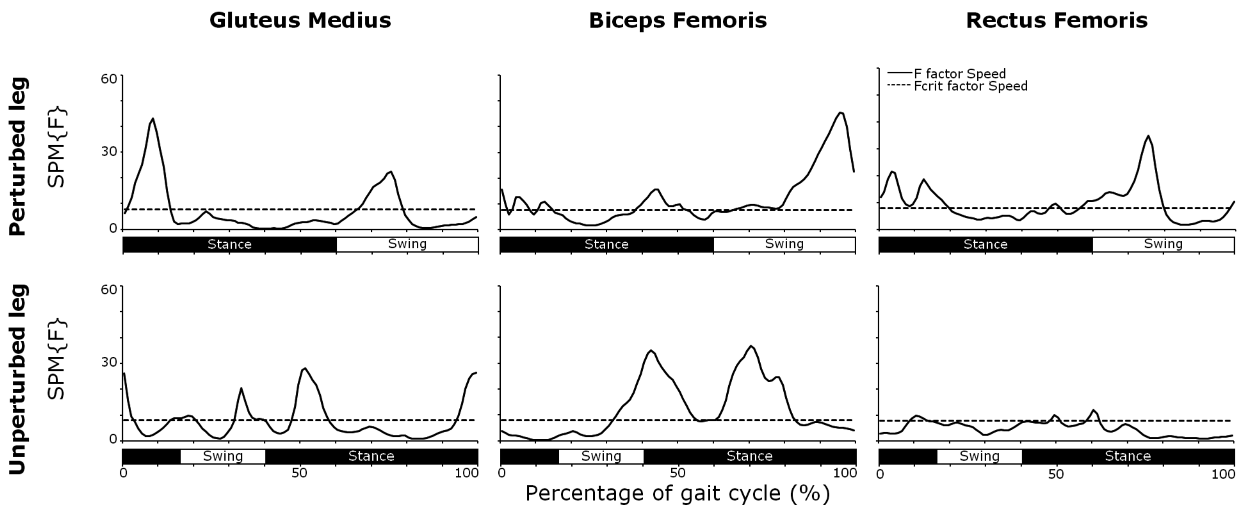

Figure A2. Main effects of Speed for Gluteus Medius, Biceps Femoris and Rectus Femoris. SPM F-values are presented for the main effect of Speed (black lines) together with the critical threshold F values (Fcrit; dashed lines). Data points that exceed the Fcrit represent the significant supra-threshold clusters.

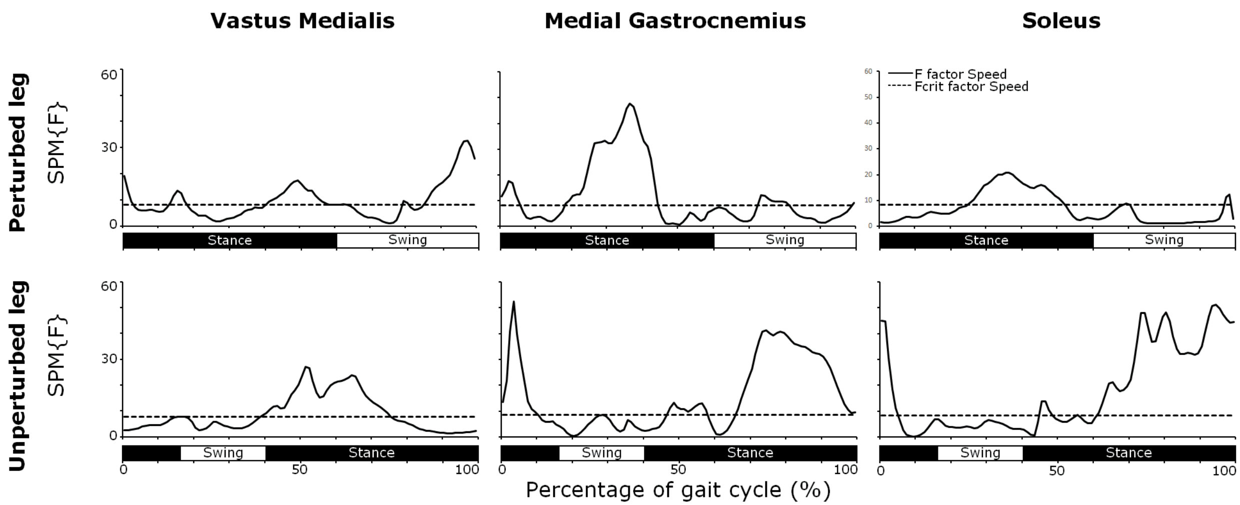

Figure A3. Main effects of Speed for Vastus Medialis, Medial Gastrocnemius and Soleus. SPM F-values are presented for the main effect of Speed (black lines) together with the critical threshold F values (Fcrit; dashed lines). Data points that exceed the Fcrit represent the significant supra-threshold clusters.

CHAPTER 5

Effects of restraining forces on propulsion and other gait characteristics during treadmill walking post-stroke

Sylvana Minkes-Weiland
Han Houdijk
Steven Floor
Paul P. Hartman
Heleen A. Reinders-Messelink
Lucas H.V. van der Woude
Rob den Otter

Clinical Biomechanics, 2025, 106664

Abstract

Background: A decreased propulsive capacity post stroke is associated with a diminished walking ability. When walking with a restraining force applied to the pelvis, more propulsion is required to enable forward progression. This may stimulate propulsion capacity in people post-stroke. Before incorporating restraining forces into training, their effects on propulsion mechanics and other gait characteristics must be evaluated. This study investigated: (1) the immediate bilateral effects of restraining forces during treadmill walking on propulsive force, braking force, and mechanical work in people post-stroke, and (2) the impact of this manipulation on step length symmetry, single support time symmetry and muscle activity. Additionally, we explored whether these effects vary with gait speed and force magnitude.

Methods: 13 Individuals post-stroke walked on a treadmill at 0.28m/s and 0.56m/s while a horizontal restraining force (0%, 5% or 10% of their body weight) was applied to the pelvis. During walking, Ground Reaction Forces and muscle activity of Gluteus Medius, Rectus Femoris, Vastus Medialis, Biceps Femoris, Tibialis Anterior, Medial Gastrocnemius and Soleus were bilaterally recorded.

Findings: Applying restraining forces up to 10% of body weight increased propulsive impulse and mechanical work while reducing braking impulse. Although no significant effects were found on step length symmetry or swing phase symmetry, subtle changes in muscle activity were observed when walking with restraining forces.

Interpretation: Restraining forces up to 10% of body weight can activate propulsive capacity. Future research should explore how this direct effect translates into long-term training effects.

Introduction

Stroke is a leading cause of long-term disability, often causing impairments in motor function that profoundly affect an individual's ability to walk. Propulsion, the positive component of the horizontal ground reaction force (GRF) during the push-off phase of the gait cycle [1], is one aspect of gait that is often severely affected after a stroke [2-5]. Improvements in propulsion have shown to go along with improvements in functional balance, walking function (i.e. walking speed and long distance walking ability) and self-perceived participation [2,6]. Therefore, training propulsive capacity could play a valuable role in post-stroke gait rehabilitation.

Gait rehabilitation after stroke facilitates recovery of the affected leg and is known to reduce dependence on the unaffected leg in generating propulsion [1]. In recent years, several studies have been conducted to assess the effectiveness of exercise interventions aimed at improving propulsive capacity [7]. However, findings remain mixed. For example, functional electrical stimulation [2], robotic training devices [8], inclined treadmill training [9] and body weight-supported treadmill training [10,11] have all been studied, with varying outcomes. Although some of these approaches have demonstrated promise, their broader application in clinical practice is often limited by the need for specialized equipment or expertise. Interestingly, exploratory studies suggest that restraining forward movement during treadmill walking, using (relatively simple) restraining horizontal forces can enhance propulsion [12, 13].

A restraining force influences propulsion by requiring the walker to generate greater forward-directed force (and mechanical power (W)) to maintain their gait speed. By applying a backward-directed force to the pelvis or legs, this approach increases the workload on the propulsive muscles. Research showed that during normal walking, the Medial Gastrocnemius and Gluteus Maximus primarily contribute to propulsion, while in more demanding conditions, the Gluteus Medius, Hamstrings, Vastus Medialis, and Soleus can provide assistance [14]. Over time, this increased effort may lead to structurally improved propulsion and thereby better functional balance, walking function and self-perceived participation [2,6], making it a potentially valuable tool in rehabilitation.

Research on restraining forces has shown that it aligns with key learning principles that may enhance its effectiveness in rehabilitation. First, achieving gains in propulsive capacity requires tapping into an unused reserve during training, referred to as 'latent propulsive capacity' [7]. Research on persons post-hemiparetic stroke showed that restraining forces can help exploit this propulsive capacity [12], suggesting restraining forces can create a potential for training effects to occur. Second, aftereffects observed in step length symmetry and gait speed during overground walking following the application of a restraining force to the leg [15] demonstrate the presence of a learning effect. This learning effect is crucial for transferring training adaptations to regular overground walking. To what the magnitude of the horizontal restraining force and imposed gait speed impact the effect, remains unclear. Specifically, it is unknown how these factors influence (1) propulsion, braking and the associated mechanical work and (2) the coordination of the gait pattern. Enhanced understanding of these effects is essential for determining whether and how restraining forces can contribute to functional gait improvements in persons post-stroke. Insights into these mechanisms will help refine the application of restraining forces in rehabilitation and maximize their effectiveness in enhancing locomotor function.

This exploratory study investigates the immediate bilateral effects of a restraining force during treadmill walking in individuals post-stroke. Specifically, we aim to determine: (1) the immediate bilateral effects of a restraining force during treadmill walking on propulsive force, braking force and the mechanical work produced by each leg to propel the body forward and maintain upright posture and (2) the effects of the manipulation on the coordination of gait in terms of step length symmetry, single support time symmetry and bilateral muscle activity. As propulsion increases with gait speed [16], these effects will be evaluated at two different speed levels to determine whether gait speed influences the response to different restraining forces.

Methods

Subjects

13 Persons suffering from stroke (Table 1) participated in this study. All participants provided written informed consent and were included if they experienced a first unilateral stroke that occurred at least 3 months before participation, had a functional ambulation score ranging from 3 to 5 and had a unilateral paresis of the leg. Participants were excluded if they suffered from visual problems, neglect, communication problems in the Dutch language, severe phatic disorders, severely impaired cognitive functions (Mini Mental State Exam score ≤ 25) or other conditions or co-morbidities that are known to affect the gait pattern. The experimental procedures were performed according to the declaration of Helsinki for medical research involving human subjects [15] and approved by a Medical

Ethical Committee (METc University Medical Center Groningen, the Netherlands; project code: 2020.111).

Table 2: Subject characteristics

Characteristic	Count	Mean ± SD	Range
Age (years)		61 ± 8.3	42-72
Weight (kg)		83 ± 15	60-106
Body length (cm)		177 ± 12	156-198
Gender (F/M)	5/8		
Months post-stroke		32 ± 12	6-53
Type stroke (ischemic/hemorrhagic)	7/6		
Location stroke (right/left hemisphere)	6/7		
Location stroke (cortical/subcortical)	7/6		
Use of walking aids during daily activities (no/yes)	11/2		
FAC score	FAC 4: 2 FAC 5: 11		4-5
Motricity index of the legs (scores ranging 0 – 100; [16])	75: 2 100: 11		75-100
Score on single leg heelrise test (affected/ unaffected leg; scores ranging 0-25; [17])		14 ± 10 / 18 ± 9	0-25 / 0-25
Spasticity measured with the Modified Ashworth Scale (yes/no; [18])	0/13		

Materials

For this study, a self-constructed pulley system was used to deliver a horizontal restraining force during treadmill walking (see fig 1; [21]). Two vertically fixed bicycle wheels guided a rope (6.14 m long from the pelvis to the first contact with the first wheel) that was fastened on one end to the subject's pelvic brim, while the other end allowed for the attachment of a load. Given this long horizontal distance, we expect that minor variations in vertical positioning due to participant height had only a minimal influence on the angle of pull. The magnitude of the restraining force depended on the load attached (0, 5 and 10% body weight (BW)). Using a pulley force, the additional average external power (W) that needs to be generated during gait is the product of treadmill belt speed and the gravitational force on the pulley mass.

Fig 1: An illustration of the experimental set-up with the instrumented treadmill, safety harness system and the pulley system from 'right to left'.

Experimental procedure

Participants walked on an instrumented treadmill (Motek, Amsterdam, The Netherlands) during six conditions of 90 seconds each. First, participants walked on a treadmill at two speeds, 0.28 m/s and 0.56 m/s, in a randomized order without any resistance. These speeds were selected to reflect functional walking capacities commonly observed in individuals post-stroke. Specifically, they fall within the range of comfortable walking speeds for this population [22], ensuring ecological validity. Subsequently, they repeated the treadmill walking with the pulley system attached to their pelvic brim, with loads of 5% and 10% of their body weight applied through the system at both speed levels (0.28 m/s and 0.56 m/s). These four load conditions (two load levels at two speeds) were presented in a randomized order. The force levels were based on earlier work applying horizontal resistance during gait [12]. The 5% BW condition was intended to apply a moderate resistive challenge, whereas the 10% BW condition served as a higher, more extreme load to explore the upper boundary of tolerability in the post-stroke population. Although the two load levels were later collapsed in the primary statistical analysis, both levels were included to allow for the inspection of potential graded effects of force magnitude. To this end, results for the 5% and 10% BW conditions were also visualized separately to inspect possible trends. While bodyweight support is known to reduce the ability to generate propulsion by decreasing the limb loading [23], it was chosen not to provide body weight support. A safety harness was effective in each of the experimental conditions. To prevent external support from affecting natural gait patterns, participants were instructed not to hold the treadmill handrails during walking.

Data collection

To assess the effects on propulsion, ground reaction forces were recorded at 100 Hz using two 3D force plates (one for each leg) that were incorporated into the treadmill. Propulsion was defined as the positive anterior ground reaction force during the push-off phase of the gait cycle [1].

To assess muscle activity, surface electromyography (EMG) was recorded bilaterally using a Trigno wireless system (Delsys Inc., Boston, MA, USA; 1925Hz) with a fixed inter-electrode distance of 10 mm of: Gluteus Medius (GM), Biceps Femoris (BF), Rectus Femoris (RF), Vastus Medialis (VM), Medial Gastrocnemius (MG), Soleus (SOL) and Tibialis Anterior (TA). Electrode locations were determined and prepared based on SENIAM guidelines [24].

Data analysis

Offline analysis of the force plate and EMG data was performed using custom-made software routines in Matlab (Matlab 2022; MathWorks, Natick, MA).

Force plate data were recursively filtered with a 15Hz second-order low-pass Butterworth filter [25] and down sampled to 100Hz for further analyses. Propulsive impulse, defined as the time integral of the positive anterior-posterior GRF during stance, and braking impulse, defined as the time integral of the negative anterior-posterior GRF, were calculated from the anterior-posterior force plate data, which was normalized to body weight per participant, and analyzed for each step. Group-averaged absolute values for propulsive impulse and braking impulse were calculated for each condition separately.

To calculate external mechanical work, defined as the work of each leg to propel the body forward and to maintain upright posture, the inverted pendulum model of walking was used [26]. This model assumes that the dynamics of the center of mass (CoM) during single stance are similar to that of an inverted pendulum and that during the step-to-step transition, the CoM is redirected from one pendular stance leg to the next [27]. In line with previous studies on treadmill walking [28] external mechanical power generated by each leg was calculated as the sum of mechanical power generated by the leg on the CoM and the power generated by the leg on the moving treadmill belt. According to equation 1

Equation 1:

$$External\ mechanical\ power = \vec{F}_{grf} \cdot \vec{V}_{CoM} + \vec{F}_{grf} \cdot \vec{V}_{belt}$$

with F_{grf} representing the GRF vector of the leg, V_{CoM} representing the velocity of the Center of Mass and V_{belt} representing the treadmill belt speed. CoM velocity was derived from GRF as described by Donelan et al (2002). Subsequently, mechanical work was calculated by integrating the mechanical power for the first double support (DS1), single support (SS) and the second double support (DS2) phase separately. Gait events were defined as the moments where the vertical GRF (GRF) on the individual force plates crossed the threshold of 50N. The work generated during DS1 is defined as collision work (negative work during collision) and work generated during DS2 as push off work (positive work during push off).

Step parameters were derived from GRF data. To assess potential temporal and spatial asymmetries in response to the applied resistance that primarily affects the swing phase, swing time and step length symmetry were included. Swing time was calculated as the time between toe-off and heelstrike of the contralateral leg. Step length, defined as the difference in anterior-posterior CoP positions at heelstrike between both legs, was calculated from the force plate data, similar to previous research [29]. Swing time symmetry and step length symmetry were calculated according to equation 2 for each step and each participant and eventually averaged for the group per condition.

Equation 2 [26]:

$$Symmetry\ index = \frac{value\ affected\ leg - value\ unaffected\ leg}{value\ affected\ leg + value\ unaffected\ leg} \times 100\%$$

The EMG data were high-pass filtered using a 10 Hz fourth order high-pass Butterworth filter to attenuate movement artefacts, full wave rectified, and low-pass filtered using a 20 Hz fourth order Butterworth filter. The EMG signals were time normalized from heelstrike to heelstrike based on the force plate data and amplitude normalized with respect to the maximal amplitude observed during all trials, for each muscle and each participant separately. To compare muscle activity between conditions, EMG amplitude was averaged within each of the four gait cycle phases; DS1, SS, DS2 and swing (SW), based on the force plate data. Group averaged activity was calculated for each muscle, walking condition and step phase separately.

Statistical analysis

Repeated measures (RM) ANOVAs using SPSS (version 23.0, IBM Corporation, Armonk, NY, USA) with the factors Force (0%, 5% and 10% of body weight), Speed (0.28m/s and 0.56m/s) and Leg (affected leg and unaffected leg) were conducted and evaluated with an alpha of 5% to assess the main effects of Force and Leg and the interactions of Force with Speed and Force with Leg on all variables. In total, 14 dependent variables were analyzed across the different conditions, including measures related to ground reaction forces, mechanical work and muscle activity. Given the exploratory nature of the study, no formal statistical tests for normality were conducted. We acknowledge potential multicollinearity, particularly among variables related to gait and muscle activity. To account for this, we performed a Repeated Measures MANOVA to examine the effects on muscle activity, which takes correlation between dependent variables into account. We first assessed significance of the multivariate test and only assessed the univariate results if the omnibus test was significant. Given the large number of outcome measures, we present a complete overview of all analyzed variables and the corresponding effects in the supplementary material. Given the abundance of data in this study, we do not assess post hoc test results for separate levels of restraining force.

This study was designed as an exploratory investigation, with the aim of broadly assessing the effects of walking with restraining forces across multiple domains of gait.

Results

All participants completed all experimental conditions, including the highest force level, without difficulty or early termination. In accordance with the aims of our study, this results section will focus on the main effects of the factors Force and Leg and the interactions of Force with Speed and Force with Leg. Table 2 and 3 of the supplementary material can be consulted for a complete overview of the statistical results and the test statistics.

Propulsive impulse, braking impulse and mechanical work

Applying restraining forces led to an increase in propulsive impulse ($F(2, 24) = 120.9$, $p = < .001$, $\eta^2_p = 0.9$) and push off work (DS2; $F(2, 24) = 12.5$, $p = < .001$, $\eta^2_p = 0.5$), while reducing braking impulse ($F(2, 24) = 120.9$, $p = < .001$, $\eta^2_p = 0.9$) and collision work (DS1; $F(2, 24) = 4.0$, $p = < .001$, $\eta^2_p = 0.3$),), as shown in the anterior-posterior GRF profile and the group averages for each condition in fig 2. A significant main effect of the factor Leg indicated differences in propulsive ($F(1, 12) = 14.7$, $p = .002$, $\eta^2_p = 0.5$) and braking impulses ($F(1, 12) = 14.7$, $p = .002$, $\eta^2_p = 0.5$) between the legs. As illustrated in Fig 2, the affected leg exhibited a lower propulsive impulse and a higher braking impulse compared to the unaffected leg.

Additionally, increases in propulsive impulse ($F(2, 24) = 3.9$, $p = .034$, $\eta^2_p = 0.2$) due to the applied forces were stronger at higher speeds while reductions in braking impulse ($F(2, 24) = 3.9$, $p = .034$, $\eta^2_p = 0.2$) were less pronounced at higher speed levels, as evidenced by a significant interaction between the Force and Speed factors. There was no difference in responses to the restraining force between the legs as no significant interaction of the Force and Leg factors was observed.

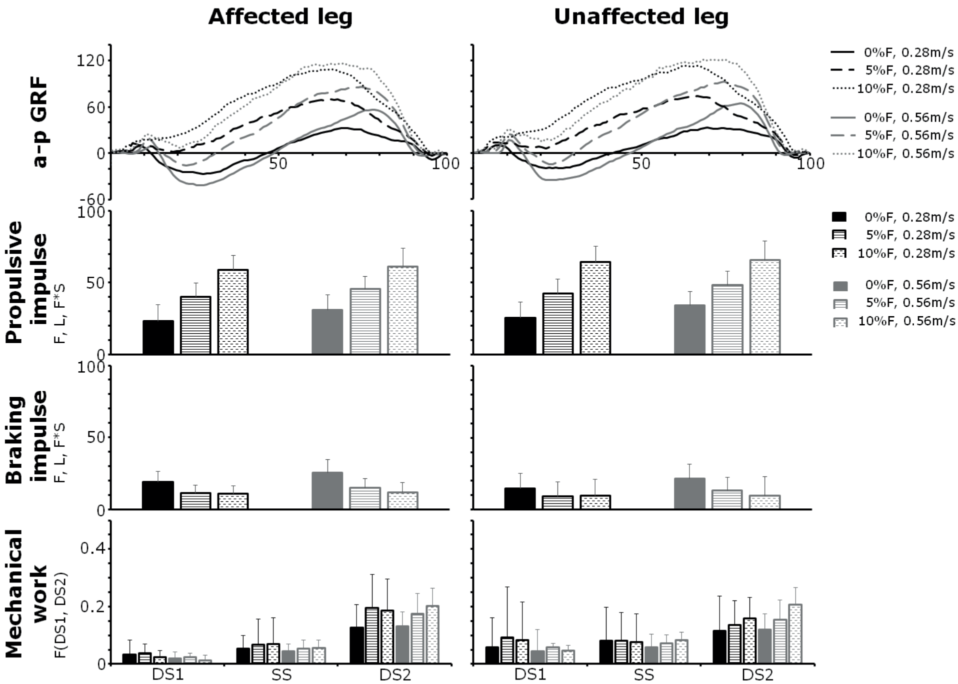

Fig. 2. Anterior-posterior GRF profiles along with group-averaged values for propulsive impulse, braking impulse, and mechanical work. The figure presents group-averaged anterior-posterior GRF (N) profiles for walking under all conditions, 0%, 5%, and 10% restraining force (F) at speeds of 0.28 and 0.56 m/s during the stance phase (0-100) as displayed on the x-axes. Additionally, the absolute average and standard deviation of propulsive impulse (N-s), braking impulse (N-s), and mechanical work (J/kg) are shown. Significant main effects for Force (F), Leg (L), and the interactions Force*Speed (F*S) and Force*Leg (F*L) are indicated on the y-axis label on the left of the figure.

Swing time symmetry and step length symmetry

Applying a restraining force had no impact on swing time symmetry or step length symmetry as there were no significant effects of the factor Force, nor an effect of the factor Leg or interactions of Force with Speed and Force with Leg.

Muscle activity

Group-averaged EMG profiles, along with the average activity per step phase, are presented in fig 3 and fig 4. As shown in these figures, applying a restraining force led to subtle increases in RF activity (DS1; $F(2, 24) = 8.7$, $p = .001$, $\eta^2_p = 0.4$ & SS; $F(2, 24) = 5.1$, $p = .014$, $\eta^2_p = 0.3$), VM activity (DS1; $F(2, 24) = 4.2$, $p = .026$, $\eta^2_p = 0.3$), TA activity (SS; $F(2, 24) = 7.4$, $p = .003$, $\eta^2_p = 0.4$ & SW; $F(2, 24) = 3.7$, $p = .040$, $\eta^2_p = 0.2$) and SOL activity (DS2; $F(2, 24) = 6.5$, $p = .005$, $\eta^2_p = 0.4$), while decreasing GM activity (DS1; $F(2, 24) = 5.5$, $p = .011$, $\eta^2_p = 0.3$, SS; $F(2, 24) = 3.6$, $p = .041$, $\eta^2_p = 0.2$, DS2; $F(2, 24) = 7.4$, $p = .003$, $\eta^2_p = 0.4$ & SW; $F(2, 24) = 12.9$, $p = < .001$, $\eta^2_p = 0.5$), MG activity (SS; $F(2, 24) = 4.1$, $p = .030$, $\eta^2_p = 0.3$) and SOL activity (DS1; $F(2, 24) = 4.5$, $p = .022$, $\eta^2_p = 0.3$), as indicated by a significant main effect of Force. There were no differences in muscle amplitudes between the legs across the four gait cycle phases, as no significant effect of the factor Leg factor was observed.

The effects of the restraining force on BF activity (DS1; $F(2, 24) = 5.4$, $p = .012$, $\eta^2_p = 0.3$) and TA activity (DS1; $F(2, 24) = 6.2$, $p = .007$, $\eta^2_p = 0.3$) varied with gait speed, as evidenced by a significant interaction effect between the factors Force and Speed. Specifically, TA activity in DS1 showed a greater increase with restraining force at lower speeds, while BF activity in DS1 increased more at higher speeds. Additionally, responses to the restraining force in RF (SS; $F(2, 24) = 4.0$, $p = .032$, $\eta^2_p = 0.3$) and TA (SS; $F(2, 24) = 3.9$, $p = .035$, $\eta^2_p = 0.2$) were more pronounced in the unaffected leg than in the affected leg, as indicated by a significant interaction between Force and Leg.

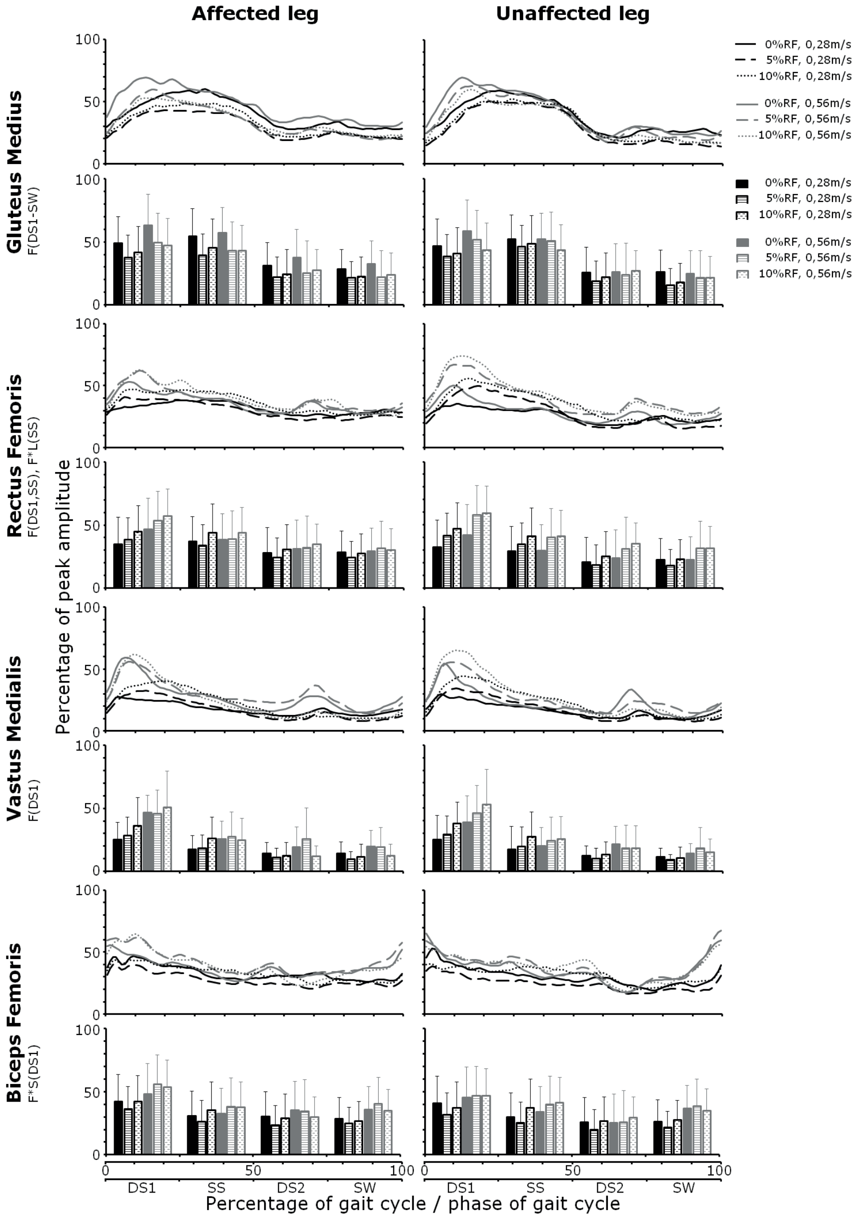

Fig 3. EMG profiles of Gluteus Medius, Rectus Femoris, Vastus Medialis and Biceps Femoris along with group-averaged activity + SD during each step phase. The figure displays group-averaged EMG profiles for walking under all conditions, 0%, 5%, and 10% restraining force (F) at speeds of 0.28 and 0.56 m/s. Additionally, the average and standard deviation for each step phase (DS1, SS, DS2, SW) are shown. Significant main effects for Force (F), Leg (L), and the interactions Force*Speed (F*S) and Force*Leg (F*L) are indicated on the y-axis label on the left of the figure.

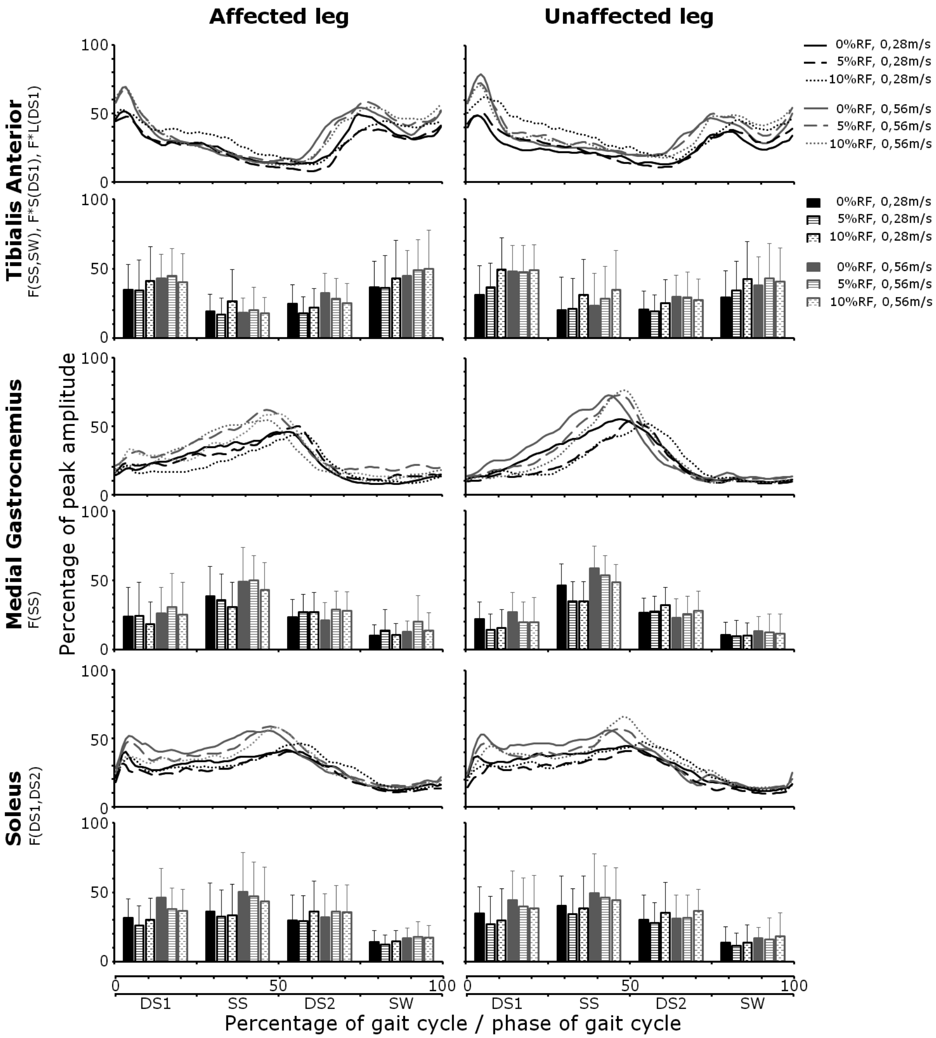

Fig 4. EMG profiles of Tibialis Anterior, Medial Gastrocnemius and Soleus along with group-averaged activity + SD during each step phase. The figure displays group-averaged EMG profiles for walking under all conditions, 0%, 5%, and 10% restraining force (F) at speeds of 0.28 and 0.56 m/s. Additionally, the average and standard deviation for each step phase (DS1, SS, DS2, SW) are shown. Significant main effects for Force (F), Leg (L), and the interactions Force*Speed (F*S) and Force*Leg (F*L) are indicated on the y-axis label on the left of the figure.

Discussion

This exploratory study examined the immediate bilateral effects of a restraining force applied to the pelvis during treadmill walking in individuals post-stroke. Specifically, we aimed to determine: (1) how restraining forces influence propulsive force, braking force, and mechanical work, and (2) how the manipulation affects gait coordination in terms of step length symmetry, single support time symmetry, and bilateral muscle activity. Effects were assessed at two different walking speeds and three different force levels. Our findings support previous research [12], confirming that restraining forces can activate propulsive capacity in persons post-stroke. Applying restraining forces resulted in an increased propulsive impulse and push-off work. This study uniquely focuses on both the mechanical and neuromuscular effects of horizontal restraining forces, distinguishing itself by examining these impacts at varying speeds and force levels.

Effects on ground reaction forces and mechanical work

When inspecting the anterior-posterior GRF data (see fig 2), both the timing and the amplitude of propulsion appear to be affected by restraining forces, suggesting that participants may adopt a different strategy for propulsion during walking. Specifically, we observe that propulsion occurs earlier in the gait cycle with restraining forces, which could indicate an adaptive change in how the participants generate propulsion. However, despite this shift in timing, mechanical work produced by the legs increases during the second double stance (DS2) phase, rather than distributing to the single stance (SS) phase. This suggests that the change in propulsion timing does not lead to a strong reorganization in temporal coordination, as the phase of mechanical work generation does not align with the shift in timing of the force impulse. If mechanical work had shifted toward SS, it could have indicated a stronger reliance on hip extensors during single stance, as is often observed in post-stroke individuals who compensate for impaired ankle push-off by increasing hip power output [1]. The absence of such a shift in mechanical work suggests that the participants did not adopt a more proximal (i.e., hip-driven) propulsion strategy, at least not within the constraints of this experimental session. Additionally, braking impulse was reduced with restraining forces, suggesting that the restraining forces contributed to mitigating the deceleration phase before impact. Although differences in collision work were observed between the restraining force conditions, the absolute values remained relatively low, indicating that the overall collision cost was limited.

Muscle activity and neuromuscular strategy

Theoretically, applying restraining forces to the pelvis increases the demand on the muscles responsible for propulsion. In general, propulsion is generated by muscular contributions at both the hip and ankle joints [6]. Strategy selection can vary among individuals, legs and level of impairment, with stroke patients often relying more on the hip due to plantarflexor weakness [1,6]. During normal walking, the Medial Gastrocnemius and Gluteus Maximus primarily contribute to propulsion, while in more demanding conditions, the Gluteus Medius, Hamstrings, Vastus Medialis, and Soleus can provide assistance [14]. Our results indicate that, under the current walking speed and restraining force conditions, only the Rectus Femoris (SS) and Soleus (DS2) showed a modest increase of activity during the generation of propulsion. The modest increase suggests a slightly elevated muscular effort in response to the restraining force but does not reflect a broader neuromuscular reorganization. Consequently, the applied manipulation did not result in a substantial reorganization of neuromuscular coordination. Aside from subtle amplitude changes in the Rectus Femoris, Soleus and a few other muscles, muscle activity remained largely unaffected in SS and DS2, indicating that the observed neuromuscular adaptations in SS and DS2 were relatively limited and did not involve shifts in the temporal structure of motor control. Interestingly, these findings contrast somewhat with previous work in healthy older adults, where horizontal impeding forces elicited more robust increases in ankle push-off intensity and associated muscle activity [13]. This difference may reflect stroke-specific impairments such as reduced selective motor control or altered neuromuscular strategies, which could limit the capacity to modulate propulsion via the ankle plantarflexors. However, we did observe changes in multiple muscles during DS1, which may suggest that participants adapted their movement strategy upon initial contact. Although posture was not directly measured, it is conceivable that participants adopted a more forward-leaning trunk and increased knee flexion, as suggested by previous similar research [31], which could in turn influence muscle activity in early stance. This adjustment may help with force absorption and prepare the body for more efficient propulsion in subsequent phases of gait.

Influence of walking speed on the responses to the restraining forces

Results indicated that speed influenced the response to the restraining force across several variables. Specifically, effects of the restraining force on propulsive impulse were stronger at higher speeds, suggesting that faster walking may amplify effects in propulsion. In contrast, reductions in braking impulse were less pronounced at higher speeds. Additionally, changes in muscle activity varied with speed showing greater increases during early stance at lower speeds in the TA and at higher speeds in the BF. The findings on propulsive and braking impulses suggest that gait speed influences how individuals adapt to restraining forces. It is also possible that more pronounced effects could be observed at even higher speeds than those tested in this study, as suggested by findings from other comparable studies in able-bodied participants [14]. The influence of speed on the response to the restraining force highlights the importance of considering walking speed when designing rehabilitation protocols, as different speeds may impact both the effectiveness and the outcomes of propulsion-focused training interventions.

Possible postural changes and their implications

The application of horizontal restraining forces may have encouraged participants to adopt a more forward-leaning posture, similar to the adaptations seen during a similar experiment [31] and uphill walking [32]. When walking uphill, individuals typically exhibit greater flexion at the hip, knee, and ankle at initial foot contact, along with a forward tilt of the pelvis and trunk [32]. These postural changes can enhance power generation, possibly explaining the earlier onset of the propulsive phase observed when walking with horizontal restraining forces. In this posture, propulsion may primarily rely on the hip [33], while ankle power may be diminished due to limited energy absorption capacity in a dorsiflexed position. In a clinical setting, a forward leaning posture may pose challenges, as some patients may feel insecure or unwilling to adopt this posture, potentially affecting their gait pattern [34]. A limitation of this study is that upper body kinematics were not recorded, making it unclear to what extent participants may have altered their posture in response to the applied restraining forces.

Feasibility, limitations and future directions

While we have assessed the direct effects of horizontal restraining forces, it remains inconclusive whether the enhanced propulsion originated from a functional neuromuscular strategy that can be transferred to normal overground walking or a dysfunctional strategy that is specific to the restraining force condition on the treadmill. Our findings do not provide evidence regarding learning effects or fatigue, nor do they indicate whether the manipulation results in lasting changes to gait patterns or walking ability. As such, no conclusions can yet be drawn about its therapeutic value. Moreover, although all participants tolerated the protocol without adverse events, the intervention was tested only in a single session. As such, the long-term feasibility, safety, and tolerability of repeated training sessions remain unknown, particularly in individuals with lower functional capacity. Our study primarily included individuals with relatively good walking ability, and we observed minimal differences between the affected and unaffected sides. From a clinical perspective, it remains to be explored whether individuals with more severe post-stroke impairments respond differently to this intervention, whether they can tolerate and sustain the applied forces, and whether the observed effects can be generalized across a broader clinical population over time. Future research should address these questions by examining long-term adaptations, transfer to overground function, therapeutic outcomes, and the clinical feasibility of implementing horizontal restraining forces in more diverse and severely affected stroke populations. Additionally, future studies should explore the presence and impact of postural changes, such as a forward lean, to determine whether such adjustments are functionally beneficial for gait rehabilitation or merely a short-term compensatory response to external forces. Given the exploratory nature and limited sample size of the current study, future work with greater statistical power is warranted to draw more definitive clinical conclusions.

Conclusions

In conclusion, this study examined the immediate bilateral effects of a restraining force during treadmill walking in individuals post-stroke. Applying a restraining force increased propulsive impulse and push-off work while reducing braking impulse. Although no significant effects were found on step length symmetry or swing phase symmetry, subtle changes in muscle activity were observed. The results suggest that restraining forces can activate propulsive capacity (in both legs) without rigorous changes in muscle activation patterns. Future research should explore long-term effects and applicability in individuals with more severe gait impairments, as well as the role of the upper body during restrained walking.

Acknowledgements The authors wish to thank the technical support of the department of Human Movement Sciences of the University Medical Center Groningen, especially Emyl Smid and Anniek Heerschop, for their technical help and advice.

Funding This work was supported by ‘Stichting Beatrixoord Noord-Nederland [project number 210.182].

References

1. Hsiao, H., Knarr, B. A., Pohlig, R. T., Higginson, J. S., & Binder-Macleod, S. A. (2016). Mechanisms used to increase peak propulsive force following 12-weeks of gait training in individuals poststroke. Journal of biomechanics, 49(3), 388-395
2. Awad, L. N., Reisman, D. S., Kesar, T. M., & Binder-Macleod, S. A. (2014). Targeting paretic propulsion to improve poststroke walking function: a preliminary study. Archives of physical medicine and rehabilitation, 95(5), 840-848.
3. Bowden, M. G., Balasubramanian, C. K., Neptune, R. R., & Kautz, S. A. (2006). Anterior-posterior ground reaction forces as a measure of paretic leg contribution in hemiparetic walking. Stroke.
4. Chen, G., Patten, C., Kothari, D. H., & Zajac, F. E. (2005). Gait differences between individuals with post-stroke hemiparesis and non-disabled controls at matched speeds. Gait & posture, 22(1), 51-56.
5. Sousa, A. S., Silva, A., Santos, R., Sousa, F., & Tavares, J. M. R. (2013). Interlimb coordination during the stance phase of gait in subjects with stroke. Archives of Physical Medicine and Rehabilitation, 94(12), 2515-2522.
6. Nadeau, S., Gravel, D., Arsenault, A. B., & Bourbonnais, D. (1999). Plantarflexor weakness as a limiting factor of gait speed in stroke subjects and the compensating role of hip flexors. Clinical biomechanics, 14(2), 125-135.
7. Alingh, J. F., Groen, B. E., Van Asseldonk, E. H. F., Geurts, A. C. H., & Weerdesteyn, V. (2020). Effectiveness of rehabilitation interventions to improve paretic propulsion in individuals with stroke–A systematic review. Clinical biomechanics, 71, 176-188.
8. Forrester, L. W., Roy, A., Hafer-Macko, C., Krebs, H. I., & Macko, R. F. (2016). Task-specific ankle robotics gait training after stroke: a randomized pilot study. Journal of neuroengineering and rehabilitation, 13, 1-6.
9. Prentice, S. D., Hasler, E. N., Groves, J. J., & Frank, J. S. (2004). Locomotor adaptations for changes in the slope of the walking surface. *Gait & posture*, *20*(3), 255-265.
10. Combs, S. A., Dugan, E. L., Ozimek, E. N., & Curtis, A. B. (2012). Effects of body-weight supported treadmill training on kinetic symmetry in persons with chronic stroke. Clinical Biomechanics, 27(9), 887-892.
11. Routson, R. L., Clark, D. J., Bowden, M. G., Kautz, S. A., & Neptune, R. R. (2013). The influence of locomotor rehabilitation on module quality and post-stroke hemiparetic walking performance. Gait & posture, 38(3), 511-517.
12. Lewek, M. D., Raiti, C., & Doty, A. (2018). The presence of a paretic propulsion reserve during gait in individuals following stroke. Neurorehabilitation and neural repair, 32(12), 1011-1019.
13. Conway, K. A., Crudup, K. L., Lewek, M. D., & Franz, J. R. (2021). Effects of horizontal impeding force gait training on older adult push-off intensity. *Medicine and science in sports and exercise*, *53*(3), 574.
14. Ellis, R. G., Sumner, B. J., & Kram, R. (2014). Muscle contributions to propulsion and braking during walking and running: insight from external force perturbations. Gait & posture, 40(4), 594-599.
15. Savin, D. N., Morton, S. M., & Whitall, J. (2014). Generalization of improved step length symmetry from treadmill to overground walking in persons with stroke and hemiparesis. Clinical Neurophysiology, 125(5), 1012-1020.

16. Peterson, C. L., Kautz, S. A., & Neptune, R. R. (2011). Braking and propulsive impulses increase with speed during accelerated and decelerated walking. Gait & posture, 33(4), 562-567.
17. World Medical Association. (2003). World Medical Association Declaration of Helsinki. Ethical principles for medical research involving human subjects. Bulletin of the world health organization, 79(4), 373.
18. Demeurisse, G., Demol, O., & Robaye, E. (1980). Motor evaluation in vascular hemiplegia. European neurology, 19(6), 382-389.
19. Hébert-Losier, K., Newsham-West, R. J., Schneiders, A. G., & Sullivan, S. J. (2009). Raising the standards of the calf-raise test: a systematic review. Journal of Science and Medicine in Sport, 12(6), 594-602.
20. Harb, A., & Kishner, S. (2023). Modified ashworth scale. In StatPearls [Internet]. StatPearls Publishing.
21. Veeger, D. H. E. J., Van der Woude, L. H., & Rozendal, R. H. (1989). The effect of rear wheel camber in manual wheelchair propulsion. J Rehabil Res Dev, 26(2), 37-46.
22. Blokland, I., Gravesteijn, A., Busse, M., Groot, F., van Bennekom, C., van Dieen, J., ... & Houdijk, H. (2021). The relationship between relative aerobic load, energy cost, and speed of walking in individuals post-stroke. *Gait & posture*, *89*, 193-199.
23. Alingh, J. F., Weerdesteyn, V., Nienhuis, B., Van Asseldonk, E. H. F., Geurts, A. C. H., & Groen, B. E. (2019). Immediate after-effects of robot-assisted gait with pelvic support or pelvic constraint on overground walking in healthy subjects. Journal of neuroengineering and rehabilitation, 16, 1-9.
24. Hermens, H. J., Freriks, B., Merletti, R., Stegeman, D., Blok, J., Rau, G., … & Hägg, G. (1999). European recommendations for surface electromyography. Roessingh research and development, 8(2), 13-54.
25. Buurke, T. J., Lamoth, C. J., Vervoort, D., van der Woude, L. H., & den Otter, R. (2018). Adaptive control of dynamic balance in human gait on a split-belt treadmill. *Journal of Experimental Biology*, *221*(13), jeb174896.
26. Kuo, A. D., Donelan, J. M., & Ruina, A. (2005). Energetic consequences of walking like an inverted pendulum: step-to-step transitions. Exercise and sport sciences reviews, 33(2), 88-97.
27. Kuo, A. D. (2007). The six determinants of gait and the inverted pendulum analogy: A dynamic walking perspective. Human movement science, 26(4), 617-656.
28. Selgrade, B. P., Thajchayapong, M., Lee, G. E., Toney, M. E., & Chang, Y. H. (2017). Changes in mechanical work during neural adaptation to asymmetric locomotion. Journal of Experimental Biology, 220(16), 2993-3000.
29. Swart, S. B., den Otter, R., & Lamoth, C. J. (2020). Anticipatory control of human gait following simulated slip exposure. Scientific Reports, 10(1), 9599.
30. Kobayashi, Y., Kubo, J., Matsubayashi, T., Matsuo, A., Kobayashi, K., & Ishii, N. (2013). Relationship between bilateral differences in single-leg jumps and asymmetry in isokinetic knee strength. Journal of applied biomechanics, 29(1), 61-67.
31. Moradian, N., Ko, M., Hurt, C. P., & Brown, D. A. (2023). Effects of backward-directed resistance on propulsive force generation during split-belt treadmill walking in non-impaired individuals. Frontiers in Human Neuroscience, 17, 1214967.
32. Leroux, A., Fung, J., & Barbeau, H. (2002). Postural adaptation to walking on inclined surfaces: I. Normal strategies. Gait & posture, 15(1), 64–74. https://doi.org/10.1016/s0966-6362(01)00181-3

33. Leteneur, S., Gillet, C., Sadeghi, H., Allard, P., & Barbier, F. (2009). Effect of trunk inclination on lower limb joint and lumbar moments in able men during the stance phase of gait. Clinical biomechanics, 24(2), 190–195. https://doi.org/10.1016/j.clinbiomech.2008.10.005
34. Boudarham, J., Roche, N., Pradon, D., Bonnyaud, C., Bensmail, D., & Zory, R. (2013). Variations in kinematics during clinical gait analysis in stroke patients. PloS one, 8(6), e66421

5

Supplemental files

Table 2: An overview of the univariate test results

	Force		Speed		Leg		Force * Speed		Force * Leg		Speed*leg		Force * Speed * Leg	
	F(2,24)	η_p^2	F(1,12)	η_p^2	F(1,12)	η_p^2	F(2,24)	η_p^2	F(2,24)	η_p^2	F(1,12)	η_p^2	F(2,24)	η_p^2
Propulsive Impulse	120.7***	0.9	24.6***	0.7	14.7*	0.5	3.9*	0.2						
Braking Impulse	120.9***	0.9	24.6***	0.7	14.7*	0.5	3.9*	0.2						
Mechanical work DS1	4.0*	0.3	7.6*	0.4										
Mechanical work SS														
Mechanical work DS2	12.5***	0.5												
Swing time symmetry			14.2*	0.5										
Step length symmetry			15.9*	0.6										

Note. This table displays the significant (*=P<0.05, **= P <0.01, ***= P <0.001) univariate F values of the repeated measures ANOVAs together with the partial eta-squared effect sizes (η_p^2).

Tabel 3: An overview of the univariate test results for muscle activity per step phase.

	Force		Speed		Leg		Force * Speed		Force * Leg		Speed*leg		Force * Speed * Leg	
	F(2,24)	η_p^2	F(1,12)	η_p^2	F(1,12)	η_p^2	F(2,24)	η_p^2	F(2,24)	η_p^2	F(1,12)	η_p^2	F(2,24)	η_p^2
Gluteus Medius														
DS1	5.5*	0.3	22.9***	0.7										
SS	3.6*	0.2												
DS2	7.4**	0.4												
SW	12.9***	0.5												
Rectus Femoris														
DS1	8.7**	0.4	13.7*	0.5										
SS	5.1*	0.3							4.0*	0.3				
DS2			8.2*	0.4										
SW			7.4*	0.4										
Vastus Medialis														
DS1	4.2*	0.3	29.6***	0.7										
SS														
DS2			7.5*	0.4										
SW			10.1**	0.5										
Biceps Femoris														
DS1			16.3**	0.6			5.4*	0.3						
SS			4.8*	0.3										
DS2														
SW			42.5***	0.8										
Tibialis Anterior														
DS1			10.5**	0.5			6.2**	0.3	3.9*	0.2				
SS	7.4**	0.4									7.7*	0.4		
DS2			19.0***	0.6										
SW	3.7*	0.2	6.5*	0.6										
Medial Gastrocnemius														

5

Tabel 3: Continued

	Force		Speed		Leg		Force * Speed		Force * Leg		Speed*leg		Force * Speed * Leg	
	$F(2,24)$	η_p^2	$F(1,12)$	η_p^2	$F(1,12)$	η_p^2	$F(2,24)$	η_p^2	$F(2,24)$	η_p^2	$F(1,12)$	η_p^2	$F(2,24)$	η_p^2
DS1			11.8**	0.5										
SS	4.1*	0.3	45.1***	0.8										
DS2														
SW			11.2**	0.5										
Soleus														
DS1	4.5*	0.3	19.1***	0.6										
SS			30.9***	0.7										
DS2	6.5**	0.4												
SW			10.5**	0.5										

Note. This table displays the significant (*=P <0.05, **= P <0.01, ***= P <0.001) univariate F values of the repeated measures ANOVAs together with the partial eta-squared effect sizes (η_p^2).

CHAPTER 6

Summary and general discussion

The work in this thesis aimed to enhance our understanding of different types of assisting or resisting technologies on muscle activation and motor control adaptations in stroke rehabilitation. The primary objectives were to (1) explore the use of different rehabilitation technologies for assisting or resisting the gait pattern to manipulate gait function (more specifically limb advancement and/or propulsion), and muscle activity related to gait function; and (2) determine the potential of these technologies for improving temporal and mechanical gait characteristics during rehabilitation post stroke. This chapter will first summarize the main findings, followed by a discussion of their scientific and clinical implications, and potential directions for future developments.

Assisting the gait pattern with the Lokomat exoskeleton

First, we examined the Lokomat, a widely used high-tech device for gait training, focusing on its role in assisting limb advancement in patients with stroke. When the Lokomat became commercially available in 2000 (Laszlo et al., 2023), exact effects of the Lokomat on the gait pattern were unknown. Over the past two decades, including the period during which research for this thesis was performed, the Lokomat has been extensively studied for its effectiveness in rehabilitation post stroke, highlighting its limitations in promoting natural gait patterns (van Kammen et al., 2020). In response, the Lokomat was updated with the FreeD module (Hocoma AG), which allows pelvic movement and mediolateral leg movements, thus promoting a more natural gait pattern and greater freedom of motion (Aurich-Schuler et al., 2019). Research on this more advanced version has demonstrated that the Lokomat can be effectively used to improve balance after stroke (Lu et al., 2021; Wu et al., 2023). However, no evidence has been found for superior improvements in other parameters, such as walking speed, when compared to conventional gait training strategies (Wu et al., 2023). Our studies with able-bodied participants and post-stroke patients contributed to the global body of knowledge on the role of the Lokomat robot system in gait function and provided valuable insights into our three main research aims.

Our study results showed that the Lokomat was inconsistent in influencing active muscular contributions. In chapter 2, consistent with previous findings (van Kammen et al., 2016; Moreno et al., 2013), we observed that in able-bodied participants, (a)symmetrical guidance could be used to influence muscle activity. Specifically, asymmetrical guidance was found to increase muscle activation in the Medial Gastrocnemius and Gluteus Medius (chapter 2). However, in patients post-stroke (chapter 3), we did not observe functional effects of (a)symmetrical guidance settings on muscle activity. Based on our findings and existing literature on the Lokomat's control mechanisms (Hidler & Wall, 2005), we conclude that the impedance control strategy does not effectively manipulate active muscular contributions during gait in persons after stroke. At maximal guidance, the Lokomat fully supports movement (Riener et al., 2010), eliminating the need for muscle activation. However, even under these conditions when no active contribution is needed, a phased activity can be observed (chapter 2 and 3). Conversely, reducing guidance allows for deviations from the reference trajectory, stimulating more autonomous control, but does not consistently lead to increased muscle activity. These findings indicate that the guidance setting, particularly in its asymmetrical form, cannot be reliably used to shift effort between the affected and unaffected leg in stroke patients. This limits the applicability of the Lokomat for manipulating task-specific loading of one leg, as explored in chapters 2 and 3.

Resisting the gait pattern by external restraining forces

This thesis demonstrated that horizontal restraining forces unilateral at the lower leg (chapter 4) or at the waist (chapter 5) can be used to manipulate gait function and muscle activity, fulfilling our first research aim. Specifically, propulsion increased in both able-bodied individuals and stroke patients as a result of walking with horizontal restraining forces (chapter 4 and chapter 5). Previous research had already shown that restraining forces applied to the trunk can enhance propulsion post-stroke (Lewek et al., 2018). However, the optimal application of these forces to maximize their effects remained unclear. This thesis contributes to a deeper understanding of how the immediate effects of restraining forces relate to muscle activity and how factors such as gait speed, the magnitude of the horizontal restraining force, and the point of application influence these effects.

In chapter 4, the immediate bilateral effects of unilateral swing leg resistance applied at the ankle on gait function during treadmill walking in able-bodied individuals were mapped. We found that propulsion increased bilaterally, while braking force decreased bilaterally with unilateral swing leg resistance. The observed changes in muscle activity likely accommodated the increased demand for progression of the trunk and the perturbed swing leg. Increased activity in the Vastus Medialis during late stance and in the Rectus Femoris throughout swing was found, likely contributing to the forward advancement of the perturbed swing leg. In addition, the increase of Medial Gastrocnemius activity likely contributed to ankle push-off for progression of the perturbed leg through swing.

In chapter 5, the immediate bilateral effects of a restraining force applied to the pelvis during treadmill walking in individuals post-stroke was investigated. We found that applying restraining forces increased propulsion, while decreasing braking. Aside from subtle amplitude changes in the Rectus Femoris, Soleus and a few other muscles, no rigorous changes were observed in muscle activity. It remains inconclusive whether the enhanced propulsion originated from a functional strategy that can be transferred to normal overground walking or a dysfunctional strategy that is specific to the restraining force condition on the treadmill.

Both chapters demonstrated that propulsion could be amplified by increasing gait speed and/or the magnitude of the horizontal force. Notably, the point of application of the horizontal restraining force differed between chapter 4 (ankle) and chapter 5 (pelvis), yet both approaches resulted in an increased propulsion in both legs. This finding adds to the current literature by showing that different points of force application can be used to enhance propulsion, providing flexibility in the potential design of gait training interventions. Moreover, no undesired biomechanical responses, such as rotational torques or asymmetry in step parameters, were observed due to the unilateral pull in chapter 4 within the current study constraints, further supporting the short-term feasibility and safety of this approach. However, it remains unclear whether individuals with more severe post-stroke impairments can tolerate or benefit from this intervention, highlighting the need for future research into its long-term effects and clinical applicability in this population.

Comparing assistive and resistive strategies for gait training

In post-stroke gait training, the concepts of assistance and resistance are used to enhance walking function. Assistance involves providing external support to facilitate movement, which can be particularly beneficial in the early stages of rehabilitation, when patients often have limited motor control or strength (Li et al., 2023). Although our studies using the Lokomat exoskeleton, that primarily provides assistance, did not yield consistent improvements in gait function (chapter 2 and 3), other robotic gait trainers that apply more adaptive or responsive assistance strategies may offer more effective support. In contrast to assistive-based training, resistance-based training applies external forces that challenge the walking pattern. This latter approach aligns with motor learning principles, suggesting that increasing task difficulty can enhance motor adaptation and lead to more permanent improvements (Maier et al., 2019). Our studies using horizontal restraining forces to resist the gait pattern (chapter 4 and 5) demonstrated clear and consistent positive direct effects across participants, highlighting the potential of this method. However, the choice between assistive and resistive gait training is thought to depend on both the patient's functional level and the specific rehabilitation goals. Assistance-based training may be more suitable for facilitating treadmill walking or achieving higher step speeds in patients with severe impairments or in earlier stages of recovery. Conversely, for patients with greater strength, resistance training may be more appropriate to increase training load, promote active participation, and target deficits such as reduced leg swing (Wu et al., 2012).

Task specific gait training: high-tech vs low-tech strategies

This thesis highlights that while high-tech strategies like the Lokomat incorporate important learning principles such as massed practice (Maier et al., 2019), which involves performing a task repeatedly in a short time period to enhance motor learning, they may not always be the most effective approach for manipulating gait function after stroke. The control algoritms in the Lokomat restricts adaptability and patient-driven control, which may hinder active motor learning. For patients with more strength, training approaches that encourage greater active participation may be more beneficial. Other high-tech robotic gait trainers, such as the Lopes II (Meuleman et al., 2015), operate according to the 'assist-as-needed' principle, providing targeted support while encouraging active patient participation (Alingh et al., 2021). These features may increase their potential to influence gait function and promote improvements in gait performance. However, although mechanical gait efficiency can be improved, Lopes training has not been shown to be superior to conventional therapy in improving gait patterns post-stroke (Alingh et al., 2021).

At the same time, low-tech strategies, such as body-weight support systems (e.g., the Andago V2.0) or simple pulley-based resistance setups, may offer more focused and flexible ways to manipulate specific aspects of gait. Though less technologically complex, these systems are often easier to understand and control, making them accessible tools for applying key motor learning principles in a straightforward way. Ultimately, the question is not whether high-tech strategies are superior, but whether it is truly necessary for a given patient or goal. This thesis demonstrates that cheap low-tech strategies can produce promising immediate effects on important training targets like propulsion. Especially in clinical settings, where cost, accessibility, and individual patient needs must be balanced, low-tech solutions may offer a more practical and powerful alternative. In this thesis it is shown that already with very simple, low tech solutions (consisting of a simple passive pully system) important adaptations in the gait pattern can be elicited.

Substitution vs restitution in gait rehabilitation

Patients recovering from a stroke often display reduced propulsive capacity (Awad et al., 2014; Bowden et al., 2006; Chen et al., 2005; Sousa et al., 2013), primarily due to impairments in calf muscle function (Nadeau et al. 1999; Olney et al., 1994). While some patients may regain ankle-driven propulsion through restitution, others may require substitution that entails a shift towards hip-driven propulsion to maintain functional walking ability. A key consideration in gait rehabilitation is whether to emphasize this restitution (aiming to restore typical movement patterns) or substitution (which focuses on compensatory strategies to optimize walking ability) approach (Kitago & Krakauer, 2013).

As the Lokomat forces walkers to follow the predefined gait pattern and compensatory behavior is strictly limited, it can be argued that the Lokomat is primarily focused on restitution of function. In contrast, walking with horizontal restraining forces allows for greater adaptability, placing more emphasis on the therapist's guidance regarding the gait pattern, individual limitations, and the patient's drive for energy efficiency (Emken et al., 2007). In this context, the extent to which a substitution or restitution strategy is pursued largely depends on the individual's capacity to adapt.

Our study data does not clearly indicate whether patients naturally adopt the most optimal propulsion strategy. Since individual capabilities and neuromuscular limitations vary, determining the most effective approach requires a careful assessment of each patient's biomechanical profile, rehabilitation phase, expectations for natural recovery, and rehabilitation goals (Rothi & Horner, 1983). Rather than rigidly pursuing either restitution or substitution, therapists should support each individual in finding their own optimal movement strategy. This requires offering sufficient freedom for exploration and adaptation, while using assistive technologies or resistance-based tools when necessary to challenge or support specific aspects of gait. In this context, both high-tech and low-tech solutions can play a role.

The role of technological advancement in sustainable rehabilitation care post stroke

Technological advancements play a vital role in ensuring that rehabilitation care remains both effective (Laut et al., 2016) and sustainable (Dellen & Labruyère, 2022; Werner et al., 2002). As healthcare demands continue to rise due to aging populations and an increasing prevalence of chronic conditions (Jane Osareme et al., 2024), innovations in rehabilitation technology have the potential to improve efficiency and support both patients and healthcare professionals. However, for these developments to be truly meaningful, it is important that they align with the needs of clinical practice and are evidence-based. The implementation of the Lokomat in the Netherlands serves as an example of how technology has sometimes been introduced in a supply-driven manner. Initially developed by the manufacturer, the Lokomat was placed in various rehabilitation centers, and only later did clinical practice and research clarify its benefits and limitations. A more effective approach may be to involve relevant stakeholders, in particular clinicians and movement scientists, much earlier in the developmental process. Clinicians, who work directly with patients, have valuable insights into practical challenges and training needs. Movement scientists, with their expertise in human motor control, biomechanics and motor learning, play a crucial role in ensuring that training principles are embedded in a proper way. Collaboration between stakeholders can guide the development of devices that are both technically sound and clinically useful. Furthermore, decision-making around the adoption of new rehabilitation technologies could benefit from a more structured, collaborative and multi-center shared approach. By bringing together expertise from different fields, investments in technology can be guided more effectively, ensuring that resources are used for innovations that best support patient care. Working together from the start may help create rehabilitation technologies that are not only advanced, but also widely applicable in clinical practice. Or as we say in Dutch: 'alleen ga je sneller, samen kom je verder!' (alone you go faster, together you go further; often attributed to Nelson Mandela).

Final conclusions

This thesis suggests that resistance-based training, particularly using low-tech solutions like horizontal restraining forces with a pulley system, could effectively stimulate propulsive function in post-stroke rehabilitation (chapter 4 and 5). These approaches can promote both substitution and restitution of normal gait by encouraging active patient participation. Given their cost-effectiveness, simplicity, and safety, these methods could be valuable in clinical settings. However, while our immediate findings along with the findings in previous research (Lewek et al., 2018; Savin et al., 2014) are promising, further research is needed to assess the long-term clinical relevance. Future studies should also examine physiological outcomes, such as cardiovascular strain, muscle fatigue, and metabolic efficiency, to better understand the broader impact of these interventions on patient health and rehabilitation potential.

References

Alingh, J. F., Fleerkotte, B. M., Groen, B. E., Rietman, J. S., Weerdesteyn, V., Van Asseldonk, E. H. F., ... & Buurke, J. H. (2021). Effect of assist-as-needed robotic gait training on the gait pattern post stroke: a randomized controlled trial. Journal of neuroengineering and rehabilitation, 18, 1-12.

Aurich-Schuler, T., Gut, A., and Labruyère, R. (2019). The FreeD module for the Lokomat facilitates a physiological movement pattern in healthy people – a proof of concept study. J. Neuroeng. Rehabil. 16:26. doi: 10.1186/s12984-019-0496-

Aurich-Schuler, T., & Labruyère, R. (2019). An increase in kinematic freedom in the lokomat is related to the ability to elicit a physiological muscle activity pattern: a secondary data analysis investigating differences between guidance force, path control, and FreeD. Frontiers in Robotics and AI, 6, 109.

Awad, L. N., Reisman, D. S., Kesar, T. M., & Binder-Macleod, S. A. (2014). Targeting paretic propulsion to improve poststroke walking function: a preliminary study. Archives of physical medicine and rehabilitation, 95(5), 840-848.

Bowden, M. G., Balasubramanian, C. K., Neptune, R. R., & Kautz, S. A. (2006). Anterior-posterior ground reaction forces as a measure of paretic leg contribution in hemiparetic walking. Stroke.

Chen, G., Patten, C., Kothari, D. H., & Zajac, F. E. (2005). Gait differences between individuals with post-stroke hemiparesis and non-disabled controls at matched speeds. Gait & posture, 22(1), 51-56.

van Dellen, F., & Labruyère, R. (2022). Settings matter: a scoping review on parameters in robot-assisted gait therapy identifies the importance of reporting standards. Journal of neuroengineering and rehabilitation, 19(1), 40.

Emken, J. L., Benitez, R., Sideris, A., Bobrow, J. E., & Reinkensmeyer, D. J. (2007). Motor adaptation as a greedy optimization of error and effort. Journal of neurophysiology, 97(6), 3997-4006.

Hocoma AG. https://www.hocoma.com/usa/us/legal-notes/.

Hidler, J. M., & Wall, A. E. (2005). Alterations in muscle activation patterns during robotic-assisted walking. Clinical Biomechanics, 20(2), 184-193.

Jane Osareme, O., Muonde, M., Maduka, C. P., Olorunsogo, T. O., & Omotayo, O. (2024). Demographic shifts and healthcare: A review of aging populations and systemic challenges. Int. J. Sci. Res. Arch, 11, 383-395.

van Kammen, K., Boonstra, A. M., van der Woude, L. H., Reinders-Messelink, H. A., & den Otter, R. (2016). The combined effects of guidance force, bodyweight support and gait speed on muscle activity during able-bodied walking in the Lokomat. Clinical biomechanics, 36, 65-73.

van Kammen, K., Boonstra, A. M., van der Woude, L. H., Visscher, C., Reinders-Messelink, H. A., & den Otter, R. (2020). Lokomat guided gait in hemiparetic stroke patients: the effects of training parameters on muscle activity and temporal symmetry. Disability and rehabilitation, 42(21), 2977-2985.

Kitago, T. O. M. O. K. O., & Krakauer, J. W. (2013). Motor learning principles for neurorehabilitation. Handbook of clinical neurology, 110, 93-103.

Laut, J., Porfiri, M., & Raghavan, P. (2016). The present and future of robotic technology in rehabilitation. Current physical medicine and rehabilitation reports, 4, 312-319.

Laszlo, C., Munari, D., Maggioni, S., Knechtle, D., Wolf, P., & De Bon, D. (2023). Feasibility of an intelligent algorithm based on an assist-as-needed controller for a robot-aided gait trainer (Lokomat) in neurological disorders: a longitudinal pilot study. Brain Sciences, 13(4), 612.

Lewek, M. D., Raiti, C., & Doty, A. (2018). The presence of a paretic propulsion reserve during gait in individuals following stroke. Neurorehabilitation and neural repair, 32(12), 1011-1019.

Li S. (2023). Stroke Recovery Is a Journey: Prediction and Potentials of Motor Recovery after a Stroke from a Practical Perspective. Life (Basel, Switzerland), 13(10), 2061. https://doi.org/10.3390/life13102061

Lu, W. A. N. G., Zheng, Y., Yini, D. A. N. G., Meiling, T. E. N. G., Zhang, X., Cheng, Y., ... & Xiao, L. U. (2021). Effects of robot-assisted training on balance function in patients with stroke: A systematic review and meta-analysis. Journal of Rehabilitation Medicine, 53(4), 2773.

Meuleman, J., van Asseldonk, E., van Oort, G., Rietman, H., & van der Kooij, H. (2015). LOPES II—design and evaluation of an admittance controlled gait training robot with shadow-leg approach. IEEE transactions on neural systems and rehabilitation engineering, 24(3), 352-363.

Maier, M., Ballester, B. R., & Verschure, P. F. (2019). Principles of neurorehabilitation after stroke based on motor learning and brain plasticity mechanisms. Frontiers in systems neuroscience, 13, 74.

Naghavi, N., Akbarzadeh, A., Tahamipour-Z, S. M., & Kardan, I. (2020). Assist-As-Needed control of a hip exoskeleton based on a novel strength index. Robotics and Autonomous Systems, 134, 103667.

Nadeau S, Gravel D, Arsenault AB, Bourbonnais D. Plantarflexor weakness as a limiting factor of gait speed in stroke subjects and the compensating role of hip flexors. Clinical Biomechanics. 1999;14(2):125-3.

Olney SJ, Griffin MP, McBride ID. Temporal, kinematic, and kinetic variables related to gait speed in subjects with hemiplegia: a regression approach. Physical therapy. 1994;74(9):872-85.

Riener, R., Lünenburger, L., Maier, I. C., Colombo, G., & Dietz, V. (2010). Locomotor Training in Subjects with Sensori-Motor Deficits: An Overview of the Robotic Gait Orthosis Lokomat. Journal of Healthcare Engineering, 1(2), 197-216.

Rothi, L. J., & Horner, J. (1983). Restitution and substitution: Two theories of recovery with application to neurobehavioral treatment. Journal of Clinical and Experimental Neuropsychology, 5(1), 73-81.

Savin, D. N., Morton, S. M., & Whitall, J. (2014). Generalization of improved step length symmetry from treadmill to overground walking in persons with stroke and hemiparesis. Clinical Neurophysiology, 125(5), 1012-1020.

Sousa, A. S., Silva, A., Santos, R., Sousa, F., & Tavares, J. M. R. (2013). Interlimb coordination during the stance phase of gait in subjects with stroke. Archives of Physical Medicine and Rehabilitation, 94(12), 2515-2522.

Werner, C., Bardeleben, A., Mauritz, K. H., Kirker, S., & Hesse, S. (2002). Treadmill training with partial body weight support and physiotherapy in stroke patients: a preliminary comparison. European Journal of Neurology, 9(6), 639-644.

Wu, M., Landry, J. M., Schmit, B. D., Hornby, T. G., & Yen, S. C. (2012). Robotic resistance treadmill training improves locomotor function in human spinal cord injury: a pilot study. Archives of physical medicine and rehabilitation, 93(5), 782-789.

Wu, L., Xu, G., & Wu, Q. (2023). The effect of the Lokomat® robotic-orthosis system on lower extremity rehabilitation in patients with stroke: a systematic review and meta-analysis. Frontiers in Neurology, 14, 1260652.

ADDENDUM

A

Nederlandse samenvatting

Dankwoord

Over de auteur

Lijst van publicaties

Nederlandse samenvatting

Een beroerte leidt vaak tot blijvende beperkingen in het looppatroon, onder meer door balansproblemen, verminderde coördinatie en een tekort aan afzetkracht. Voor een doeltreffende looprevalidatie zijn trainingsmethoden nodig die taak-specifiek zijn en gericht bijdragen aan het verbeteren van het looppatroon. In dit proefschrift is onderzocht in welke mate revalidatietechnologieën die ondersteuning of weerstand bieden tijdens het lopen, kunnen bijdragen aan herstel van loopvaardigheid na een beroerte. Hiertoe zijn verschillende onderzoeken uitgevoerd bij zowel gezonde volwassenen als bij personen die een beroerte hebben gehad. De centrale doelstellingen waren:

1. Evalueren hoe ondersteuning of weerstand kan worden toegepast om specifieke aspecten van het looppatroon, in het bijzonder de beenzwaai, afzet en bijbehorende spieractiviteit, te beïnvloeden;
2. Bepalen welk potentieel deze technologieën hebben voor het veranderen van temporele en biomechanische eigenschappen van het lopen tijdens revalidatie na een beroerte.

Ondersteunende technologie: het Lokomat exoskelet

In dit proefschrift (**hoofdstuk 2 en hoofdstuk 3**) is het effect van de Lokomat, een veelgebruikt robotisch hulpmiddel voor looptraining, onderzocht. Specifiek is gekeken naar het effect van bewegingsondersteuning op de spieractiviteit.

Onze studies met zowel gezonde proefpersonen (**hoofdstuk 2**) als met mensen na een beroerte (**hoofdstuk 3**) tonen aan dat de Lokomat slechts beperkt de spieractiviteit kan beïnvloeden. Bij gezonde deelnemers kon (a)symmetrische ondersteuning de activiteit van onder andere de Mediale Gastrocnemius en Gluteus Medius verhogen, maar bij mensen na een beroerte bleven dergelijke functionele effecten uit. Een mogelijke verklaring hiervoor is dat de Lokomat werkt met impedantiecontrole, een controle mechanisme waarbij de mate van ondersteuning afhankelijk is van de afwijking ten opzichte van het vooraf ingestelde looppatroon. Deze vorm van controle lijkt de actieve spierbijdrage niet effectief te beïnvloeden. Bij maximale bewegingsondersteuning neemt de Lokomat de beweging volledig over, terwijl bij verminderde bewegingsondersteuning de spieractiviteit niet evenwichtig toeneemt. Hierdoor kan deze instelling, ook in asymmetrische vorm, niet doeltreffend worden ingezet om de spieractiviteit in het aangedane en niet-aangedane been te beïnvloeden. Dit beperkt de toepasbaarheid van de Lokomat voor het gericht manipuleren van spieractiviteit bij de bestudeerde proefpersoon groepen, zoals onderzocht in **hoofdstukken 2 en 3**.

Weerstand biedende technologie: horizontale trekkrachten

Dit proefschrift laat zien dat horizontale weerstand biedende krachten, via een pulleysysteem toegepast bij de enkel (**hoofdstuk 4**) of midden op het bekken (**hoofdstuk 5**), het looppatroon en de spieractiviteit kunnen beïnvloeden. Zowel bij gezonde deelnemers als bij zelfstandig lopende mensen na een beroerte nam de afzetkracht toe. De effecten bleken afhankelijk van loopsnelheid, de grootte van de kracht en de plaats (enkel/bekken) van aangrijpen van de weerstand biedende trekkracht.

In **hoofdstuk 4** werd aangetoond dat een weerstand biedende trekkracht bij de rechter enkel leidde tot meer afzetkracht en minder remkracht tijdens lopen op de loopband in beide benen. Dit ging gepaard met verhoogde activiteit van de Vastus Medialis tijdens de late standfase en van de Rectus Femoris gedurende de zwaaifase, wat waarschijnlijk bijdroeg aan de voorwaartse verplaatsing van het verstoorde zwaaibeen. Daarnaast droeg de verhoogde activiteit van de Mediale Gastrocnemius vermoedelijk bij aan de enkelafzet voor de voortbeweging van het verstoorde been tijdens de zwaaifase.

In **hoofdstuk 5** werd bij mensen na een beroerte een weerstand biedende horizontale trekkracht bij het bekken onderzocht. Ook hier nam beiderzijds de afzet toe en de remkracht af, hoewel veranderingen in spieractiviteit beperkt bleven tot subtiele veranderingen in onder andere de Rectus Femoris en Soleus.

Beide studies tonen aan dat verhoging van de loopsnelheid of grootte van de weerstand de afzet verder versterkt, en dat verschillende aangrijpingspunten (enkel of bekken) vergelijkbare effecten op de afzetkracht kunnen opleveren. Daarbij werden geen ongewenste biomechanische bijwerkingen, zoals rotatiekrachten of asymmetrie, gevonden. Hoewel de effecten op afzetkracht kunnen duiden op een relevante trainingsprikkel voor het verbeteren van het looppatroon, blijft het nog onduidelijk of deze effecten overdraagbaar zijn naar normaal lopen buiten de loopbandcontext. Daarnaast vraagt de toepasbaarheid bij ernstiger aangedane patiënten en de langetermijneffecten om nader onderzoek.

Vergelijking van technologieën en klinische implicaties

In de looprevalidatie na een beroerte worden zowel ondersteunende als weerstand biedende technologieën toegepast. Ondersteunende technologieën, zoals de Lokomat, bieden bewegingsondersteuning om het lopen te faciliteren en zijn met name relevant in de vroege revalidatiefase wanneer kracht en motorische controle beperkt zijn. Hoewel de Lokomat veel herhalingen mogelijk maakt, lieten onze studies (hoofdstuk 2 en 3) geen effectieve beïnvloeding van spieractiviteit zien. Mogelijk bieden adaptievere robotsystemen, die werken volgens het 'assist-as-needed' principe, meer potentieel.

In tegenstelling tot ondersteunende technologieën verhoogt het opleggen van een weerstand biedende kracht tijdens het lopen de taakuitdaging. Deze benadering sluit aan bij principes van motorisch leren, waarbij een hogere taakmoeilijkheid de motorische adaptatie kan bevorderen en kan leiden tot duurzamere verbeteringen. De studies met horizontale weerstand biedende krachten (hoofdstuk 4 en 5) toonden duidelijke en consistente effecten op spieractivatie en afzetkracht, wat de potentie van deze technologie onderstreept.

High-tech versus low-tech

Hoewel high-tech robotica zoals de Lokomat geavanceerde mogelijkheden bieden, beperken hun algoritmen vaak de aanpasbaarheid en actieve inzet van de loper. Low-tech oplossingen, zoals eenvoudige katrolsystemen zijn daarentegen betaalbaar en gemakkelijk toepasbaar. Deze low-tech technologieën kunnen specifieke aspecten van het looppatroon effectief beïnvloeden en blijken in staat onmiddellijke aanpassingen in afzetkracht te bewerkstelligen.

Restitutie versus substitutie

Een centrale vraag in looprevalidatie is of de nadruk moet liggen op restitutie (focus op het herstellen van beweegpatronen) of substitutie (focus op het vinden van compensatie strategieën om het looppatroon te optimaliseren). Terwijl de Lokomat primair restitutie nastreeft door een vast looppatroon op te leggen, laat een weerstand biedende kracht tijdens het lopen meer ruimte voor variatie en kan daarmee zowel restitutie als substitutie nagestreefd worden afhankelijk van de capaciteit van de loper.

Aangezien individuele mogelijkheden en neuromusculaire beperkingen na een beroerte sterk variëren, vraagt het bepalen van de meest effectieve aanpak om een zorgvuldige beoordeling van het biomechanisch profiel van de patiënt, de revalidatiefase, de verwachtingen ten aanzien van natuurlijk herstel en de revalidatiedoelen. In plaats van te kiezen voor óf restitutie óf substitutie, zouden therapeuten iedere patiënt moeten ondersteunen in het vinden van een optimale bewegingsstrategie. Ondersteunende technologieën of weerstand biedende technologieën kunnen waar nodig worden ingezet om specifieke aspecten van het looppatroon te prikkelen of te ondersteunen. In deze context kunnen zowel high-tech als low-tech oplossingen een waardevolle rol spelen.

Conclusie

Dit proefschrift laat zien dat weerstand biedende krachten tijdens het lopen veelbelovend zijn voor het stimuleren van afzetkracht bij mensen na een beroerte. Deze low-tech technologieën zijn kosteneffectief, veilig en eenvoudig inzetbaar, en kunnen zowel restitutie als substitutie ondersteunen door actieve deelname te bevorderen. Verder onderzoek is echter nodig om de lange termijn effecten en de klinische relevantie te beoordelen. Toekomstige studies zouden daarnaast fysiologische uitkomsten moeten onderzoeken, zoals cardiovasculaire belasting, spiervermoeidheid en metabole efficiëntie, om beter inzicht te krijgen in de bredere impact van deze interventies op het loopvermogen en de gezondheid van de patiënt, en het revalidatiepotentieel.

Dankwoord

Hier zijn we dan: aangekomen op een van de laatste bladzijden van dit proefschrift. Er zijn tijden geweest dat ik niet had verwacht dat ik ooit aan een proefschrift zou beginnen. Later waren er momenten dat ik twijfelde of ik deze mijlpaal wel zou bereiken, of dat ik verlangde dat het einde al in zicht was. Maar nu is het zover: er ligt een, zo durf ik inmiddels zelf te zeggen, prachtig boekje voor me. Dit was nooit gelukt zonder de aanmoediging, steun, kritische blik en het vertrouwen van velen om mij heen. Aan hen draag ik dit dankwoord op.

Han, jij sloot je in juni 2020 aan bij het promotieteam. Daarvoor hadden we elkaar al eens gezien op congressen, maar verder kenden we elkaar nog niet goed. Je kreeg een aanstelling bij Bewegingswetenschappen in Groningen en jouw expertise sloot mooi aan bij het thema van mijn proefschrift. Dat is in de afgelopen jaren duidelijk gebleken. Je hebt inhoudelijk veel toegevoegd aan de laatste twee artikelen en me tijdens onze afspraken vaak aan het denken gezet over biomechanische vraagstukken. Omdat je je midden in de coronaperiode aansloot, waren fysieke bijeenkomsten lastig. Onze overleggen vonden meestal online plaats, en omdat ik kort daarna bij Revalidatie Friesland begon en mijn promotie parttime vanuit huis afrondde, zagen we elkaar niet eens zo vaak in persoon. Toch bleek je hierin heel flexibel. Je gaf me het gevoel dat je deur altijd (digitaal) openstond. Wanneer ik je nodig had, wist ik je te vinden en dacht je actief mee. Toen ik het lastig vond om mijn promotie te combineren met mijn werk bij Revalidatie Friesland en mijn gezinsleven, hield jij me scherp. Dankzij jouw hulp, onder andere bij het maken van een duidelijke planning van werkzaamheden en het inplannen van overleg- en feedbackmomenten rondom de schrijfweekenden van de Graduate School, ben ik uiteindelijk over de eindstreep gekomen. Ook bij de praktische zaken rondom de promotie trad je snel en daadkrachtig op. Nu staan we aan het eind van onze samenwerking aan dit proefschrift, maar hopelijk aan het begin van een nieuwe samenwerking op een ander niveau. We blijven elkaar zien op congressen en ik kijk uit naar verdere samenwerking tussen Bewegingswetenschappen Groningen en Revalidatie Friesland. Bedankt Han!

Luc, aan het begin van het traject was jij mijn eerste promotor. Later, vanwege je pensionering, droeg je deze rol over aan Han en werd je geregistreerd als tweede promotor. Ik herinner me nog goed dat ik je leerde kennen als docent van een van de vakken tijdens mijn studie Bewegingswetenschappen. Als verlegen student was ik altijd onder de indruk van je: een wijze man, druk met allerlei zaken. Toch bleek, toen Rob en ik bij je aanklopten met een idee voor een Master-PhD, dat je heel bereikbaar was. Je deur stond letterlijk altijd open. Je was oprecht geïnteresseerd, had altijd het beste met me voor en was bereid je volledig in te zetten. Toen ik door de pandemie vertraging opliep en een jaar verlenging kreeg, steunde je me. Ook gaf je me ruimte om andere ervaringen op te doen tijdens mijn promotietraject. Zo steunde je mijn betrokkenheid bij de organisatie van een symposium en bij het behalen van mijn basiskwalificatie onderwijs. Daarnaast kon ik dankzij jou ervaring opdoen met onderwijs geven bij Bewegingswetenschappen. Hoewel formele regels en beurzen soms beperkend waren, was jij bereid hier tegenin te gaan. Toen er tijdens mijn promotietraject een vacature vrijkwam bij Revalidatie Friesland, belde ik jou om te vragen of je erachter stond dat ik mijn promotie parttime naast die functie zou afronden. Jij had altijd vertrouwen: in mij, in mijn keuzes en in een goede afronding van mijn proefschrift. Je gaf me het gevoel dat ik het goed deed (ook al twijfelde ik daar zelf soms over). Tijdens het proefschrift benadrukte je steeds het belang van het effect van onze experimenten op vermogen. Daar moest volgens jou altijd iets van terugkomen. Daarom haal ik het hier ook aan, zij het in een andere context: jij hebt een ongelofelijk groot vermogen om je in te zetten voor je promovendi, om kansen te creëren en om te geloven in anderen. Bedankt dat je ook na je pensionering zo betrokken bent gebleven en me zoveel mogelijkheden hebt gegeven. Zonder jou had ik hier nu niet gestaan. Bedankt Luc!

Ja, **Rob**, dan ben jij. Wij gaan al heel wat jaren terug. Als bachelor student mailde ik je naar aanleiding van een advertentie die ik had zien hangen op het prikbord van Bewegingswetenschappen. Je zocht samen met Klaske van Kammen een student die kon helpen bij onderzoeksmetingen met de Lokomat bij Revalidatie Friesland. Ik wilde graag ontdekken of onderzoek iets voor mij was. Dat deze mogelijkheid zich voordeed in Beetsterzwaag, mijn eigen dorp, kon ik natuurlijk niet laten schieten. Hoewel de advertentie zich richtte op masterstudenten, vond je het geen probleem dat ik 'nog maar' een bachelor student was. We spraken elkaar al snel, en dat was het begin van onze samenwerking, die inmiddels zo'n tien jaar duurt. Samen bedachten we een plan voor een master-PhD: een promotietraject dat we inhoudelijk binnen het team vormgaven en dat zelfs al tijdens mijn studie Bewegingswetenschappen van start ging. Mijn bachelor afstudeerproject resulteerde in het eerste artikel van dit proefschrift, en mijn master afstudeerproject in het tweede. Tijdens mijn promotietraject was jij mijn eerste begeleider. Ik herinner me de vele momenten in jouw kamer, waar we samen het whiteboard volkladden met ideeën, vaak met humor en een kop koffie erbij (Ik zie mezelf nog zitten met die rode koffiemok zonder oortje. Heb je die eigenlijk nog?). Later, toen fysiek afspreken moeilijker werd door corona of mijn werk in Beetsterzwaag, spraken we elkaar vaak via Skype en later via Teams. Al liep Skype met jou meestal soepeler dan Teams. Na elk overleg kon ik weer verder met een lijst aan nieuwe ideeën. Bedankt voor de jarenlange samenwerking!

Met het drukken van dit proefschrift eindigt onze samenwerking gelukkig niet. Inmiddels begeleiden we samen studenten Bewegingswetenschappen bij hun onderzoeksprojecten bij Revalidatie Friesland. Ik kijk uit naar een mooi vervolg van onze samenwerking, bedankt Rob!

Heleen, wij leerden elkaar al voor mijn promotietraject kennen. Toen ik als bachelor student startte met een onderzoeksstage bij Revalidatie Friesland, ontving jij me daar, samen met Klaske van Kammen, met veel enthousiasme. Onze samenwerking is in de loop der jaren steeds verder geïntensiveerd. Vooral in de laatste jaren, waarin we collega's werden bij Revalidatie Friesland, hebben we veel contact gehad. Je hebt ontzettend veel voor me betekend, op meerdere vlakken. Hoewel jij je inhoudelijke bijdrage aan het proefschrift nooit zo groot vond, zie ik dat als een duidelijke onderschatting. Je bijdrage was juist van grote waarde, en je psychologische steun misschien nog wel belangrijker. Ik kon al mijn gedachten en gevoelens bij je kwijt, en al pratend kwamen we vaak samen tot een goed inhoudelijk of praktisch plan. Je hebt me naar de eindstreep geholpen door me te laten durven kiezen voor wat ik belangrijk vond. Soms zette je zelfs de eerste stap voor mij, zodat ik niet langer hoefde te twijfelen over hoe iets aan te pakken. Jij kent me goed en door jou heb ik ook mezelf beter leren kennen. Daar ben ik je ontzettend dankbaar voor. Ook nu, na mijn promotietraject, blijven we intensief samenwerken bij Revalidatie Friesland. Jij wees me ruim drieënhalf jaar geleden op de vacature voor onderzoeker, en dankzij jou ben ik nu precies waar ik wil zijn: onderzoeker met de blik op de revalidatiepraktijk. Ook nu we elkaar op kantoor zien ben je nog even enthousiast als die eerste keer. Ik waardeer onze samenwerking ontzettend, bedankt dat je altijd open staat om samen te sparren. Ik ben ervan overtuigd dat we nog een mooie samenwerking tegemoet gaan als bewegingswetenschappers bij Revalidatie Friesland!

Tijdens het bedenken en uitvoeren van de onderzoeken had ik gelukkig ook hulp van mensen uit de praktijk. Zonder deze belangrijke partners was het onderzoek met patiënten nooit van de grond gekomen. Er was op elk moment een revalidatiearts betrokken bij het PhD-team. In het begin was dit **Annemarijke Boonstra**, die later het stokje overgaf aan **Paul Pieter Hartman**. Ondanks jullie drukke agenda's kon ik jullie altijd bereiken en leverden jullie een actieve bijdrage. Heel erg bedankt! Ook andere collega's uit de praktijk waren bereid actief mee te denken. In het bijzonder wil ik **Steven Floor** noemen, die niet alleen meedacht, maar ook een grote drijvende kracht was bij de inclusie van patiënten voor het laatste onderzoek en zich actief inzette tijdens de metingen.

Ook **bachelor- en masterstudenten bewegingswetenschappen** hebben op verschillende momenten meegewerkt aan de dataverzameling voor de onderzoeken in dit proefschrift. Ik vond het altijd erg leuk om met jullie samen te werken en jullie te begeleiden bij academische opdrachten of afstudeerprojecten. Bedankt voor jullie inzet en de fijne samenwerking!

Daarnaast wil ik mijn oud-collega's van de 'PhD-gang' van Bewegingswetenschappen bedanken voor de inspirerende gesprekken, gedeelde frustraties en het plezier tijdens de gezamenlijke koffiepauzes. In het bijzonder wil ik **Bregje** en **Sjoukje** noemen, met wie ik tijdens de corona lockdowns de koffiepauzes naar Skype verplaatste. Zo hielden we elkaar niet alleen op de hoogte van onze PhD-projecten, maar ook van ons leven buiten het werk. Ook wil ik graag **Iris** noemen. Als ik chocoladekoekjes zie, denk ik nog altijd aan jou. We deelden onze frustraties en hielp elkaar daarmee telkens weer vooruit. Bedankt dat je altijd bereikbaar bent, ook al zien we elkaar nu veel minder. **Anniek**, we hadden veel contact toen ik weer data mocht verzamelen na de grootste coronapiek. Je hielp me niet alleen met mijn onderzoek, maar zorgde ook altijd voor gezelligheid. Bedankt daarvoor!

Ook mijn **huidige collega's van Revalidatie Friesland** wil ik niet vergeten, in het bijzonder de collega's van de Research Commissie, de Revalidatie Friesland Academie en het gangbeeldteam. Jullie leefden de afgelopen periode met me mee en bouwden een feestje voor me toen het akkoord van de beoordelingscommissie binnenkwam. Gedurende het traject waren jullie er om tegenslagen te bespreken, successen te vieren en ervaringen te delen. Bedankt voor jullie interesse, medeleven en waardevolle adviezen!

Ook wil ik mijn waardering uitspreken voor de samenwerking met Revalidatie Friesland. **Peter Tammeling**, naarmate mijn promotietraject vorderde en ik bij Revalidatie Friesland in dienst kwam als onderzoeker, kregen wij steeds meer contact. Ik wil je bedanken voor de ondersteuning die je mij gaf bij het afronden van mijn PhD. Zonder jouw steun had ik deze mijlpaal nog niet kunnen bereiken. Daarnaast heb je me prachtige kansen geboden om mijn expertise in loopvaardigheid verder uit te bouwen, onder andere via cursussen en werkzaamheden als gangbeeldanalist. Ik kijk uit naar het opzetten van verder onderzoek binnen het thema houding en beweging bij Revalidatie Friesland en ben dankbaar voor het waardevolle werk dat ik hier mag doen!

Mijn dank gaat uit naar iedereen die op welke wijze dan ook heeft bijgedragen aan de onderzoeken in dit proefschrift: **alle coauteurs, studenten, labbeheerders,**

secretaresses, planners, patiënten en proefpersonen. Zonder jullie inzet zouden deze onderzoeken niet mogelijk zijn geweest. Hartelijk dank!

Daarnaast wil ik ook de ondersteuning van **subsidieverstrekkers** noemen. Voor het onderzoek in hoofdstuk 3 ontvingen we subsidie van het Revalidatiefonds (inmiddels HandicapNL), en voor het onderzoek in hoofdstuk 4 van Stichting Beatrixoord.

En dan mijn eigen persoonlijke omgeving.

Hé **pap & mam**, 'Wat nou als het lukt'. Ik denk dat dit gelukt is, en dat is mede dankzij jullie. Jullie hebben mij altijd gestimuleerd om hard te werken. We zien elkaar vaak, en daardoor zijn jullie altijd op de hoogte van wat er speelt. Jullie hebben alles meegemaakt vanaf die ondersteunende zijlijn. Jullie boden een luisterend oor (en vaak ook advies), en hielpen me daarnaast ontzettend aan de praktische kant. Jullie passen vaak op onze kinderen. Ik ben blij dat onze meisjes dezelfde opvoeding van jullie meekrijgen als die ik heb gehad, al zijn jullie wel wat 'losser' geworden… Jullie onvoorwaardelijke steun heeft me gebracht waar ik nu ben. Dank jullie wel voor alles!

Mariëlle, mijn zus. Of zoals bij ons thuis veelvuldig klinkt 'tante'. We hebben een hechte band en doen samen met de meisjes veel leuke dingen. Je hebt het beste met ons voor en dat uit zich vooral in het verwennen met aandacht, liefde en speelgoed van de meisjes. Bedankt voor dat je je altijd om ons bekommert en aan ons denkt!

Roy, mijn broertje. Wij waren de eerste twee in de familie die aan de universiteit gingen studeren, en het gesprek over het dubbeltje en het kwartje is daarbij meermaals gevallen. Hoewel we ons studiepad allebei anders aanvlogen, zijn we allebei op een mooi punt gekomen. Fijn dat we onze hoogtepunten nog altijd met elkaar kunnen delen!

En dan, **Mats**, jij was mijn rots in de branding. Jij was erbij vanaf het begin en stimuleerde me al tijdens mijn bachelor om te ontdekken wat onderzoek precies inhield. Je steunde me bij de volgende stappen in het onderzoek en hielp me zowel praktisch als inhoudelijk waar je kon. Terwijl mijn proefschrift groeide, groeide onze relatie ook. We gingen samenwonen, eerst in Groningen en later in Beetsterzwaag, waar we inmiddels een mooi huis hebben gekocht en verbouwd. We trouwden en kregen twee prachtige meisjes: Fabiënne en Mylène. Jij bracht ontelbare keren de kinderen op bed, zodat ik 's avonds nog even achter mijn laptop kon zitten. Je gaf me de ruimte om tijd en energie in mijn proefschrift te steken en bleef me tegelijkertijd stimuleren. Soms dacht je inhoudelijk mee, met je achtergrond als bewegingswetenschapper en fysiotherapeut, en hield je me scherp. Bedankt voor je eindeloze geduld, liefde en vertrouwen. Dit proefschrift is ook een beetje van jou. Ook jouw ouders en omi wil ik hartelijk bedanken. Jullie laten me altijd voelen dat ik echt deel uitmaak van de familie. Bedankt!

Mats, je bent samen met **Sara** mijn paranimf. Ook Sara ken ik al van voordat ik aan mijn PhD-traject begon. We werkten samen tijdens onze studies bij de bakker, waar we lief en leed deelden. En dat doen we, zoveel jaar later, nog steeds. Een date bij Rinsma en ik was een afwijzing van een tijdschrift voor even vergeten. Bedankt dat we nog altijd de goede én de slechte momenten met elkaar delen. Je bent een fijne vriendin die me bijstaat, en ik ben je dankbaar dat je dit ook tijdens de promotieplechtigheid als paranimf zal doen.

Aan iedereen die op welke manier dan ook heeft bijgedragen aan dit proefschrift, of jullie nu hierboven genoemd zijn of niet: **hartelijk dank!**

Over de auteur

Sylvana Minkes-Weiland werd op 21 oktober 1994 geboren in Leeuwarden. Op jonge leeftijd verhuisde zij naar Beetsterzwaag, waar zij, op een kort uitstapje naar Groningen na, nog altijd woont. Na het behalen van haar VWO-diploma begon zij in 2013 aan de studie Bewegingswetenschappen aan de Rijksuniversiteit Groningen. In 2016 vervolgde zij haar opleiding met de master Human Movement Sciences, met als specialisatie Revalidatie.

Tijdens haar bachelor raakte Sylvana betrokken bij onderzoek naar de Lokomat looprobot bij Revalidatie Friesland. Dit resulteerde in 2018 in haar eerste wetenschappelijke publicatie, tevens het jaar van afstuderen. Aansluitend aan de master startte Sylvana een promotietraject, een samenwerking tussen de afdeling Bewegingswetenschappen (UMCG/RUG) en Revalidatie Friesland. Tijdens haar PhD behaalde zij in 2020 de Basiskwalificatie Onderwijs en verdiepte zij haar expertise in loopvaardigheid door aanvullende scholing en werkzaamheden op het gebied van gangbeeldanalyse. Sinds 2022 is zij werkzaam als onderzoeker bij Revalidatie Friesland.

In 2020 trouwde Sylvana met Martsen Minkes. Samen hebben zij twee dochters: Fabiënne (2021) en Mylène (2023).

Lijst van publicaties

Weiland, S., Smit, I. H., Reinders-Messelink, H., van der Woude, L. H., van Kammen, K., & den Otter, R. (2018). The effect of asymmetric movement support on muscle activity during Lokomat guided gait in able-bodied individuals. *PLoS One*, *13*(6), e0198473.

Minkes-Weiland, S., Reinders-Messelink, H. A., Boonstra, A. M., van der Woude, L. H., & den Otter, R. (2022). Effects of asymmetrical support on lower limb muscle activity during Lokomat guided gait in persons with a chronic stroke: an explorative study. European Journal of physical and rehabilitation Medicine, 58(5), 693.

Minkes-Weiland, S., Houdijk, H., Floor, S., Hartman, P. P., Reinders-Messelink, H. A., van der Woude, L. H. V., & den Otter, A. R. (2025). Effects of restraining forces on propulsion and other gait characteristics during treadmill walking post-stroke. *Clinical Biomechanics*, 106664.

Minkes-Weiland, S., Houdijk, H., Reinders-Messelink, H. A., van der Woude, L. H. V., Hartman, P. P., & den Otter, A. R. Effects of unilateral swing leg resistance on propulsion and other gait characteristics during treadmill walking in able-bodied individuals. *Biomechanics,* 5(4), 71.

Gotink, C. B., **Minkes-Weiland, S**., Fransen, I. E. C., Reinders-Messelink, H., Heesink, L., & Tabak, M. (2025). Self-regulation in eHealth: definition, contributing factors, and experiences from blended rehabilitation care. *Disability and rehabilitation*, 1-10.

van der Weerd, L. P., Hurkmans, J., Reinders, J. J., **Minkes-Weiland, S**., Woldring, C., Drenth, H., & Finnema, E. (2025). Facilitators and barriers to interprofessional learning, collaboration, and identity development during interprofessional internship in rehabilitation care: A qualitative study. *Journal of interprofessional care*, 1-14.