Quadriceps-hamstrings muscle co-activation during the swing phase of walking is modulated by task constraints in healthy adults

Ellis A.M. Van Can[1,2], Han Houdijk[1], Tom J.W. Buurke[1,2,*]

[1]University of Groningen, University Medical Center Groningen, Department of Human Movement Sciences, Groningen, The Netherlands

[2]KU Leuven, Department of Movement Sciences, Leuven, Belgium

*Corresponding author

Address: Antonius Deusinglaan 1, 9713 AV, Groningen, The Netherlands

Tel: +31 50 361 6015

Email: t.j.w.buurke@umcg.nl

Keywords: Electromyography, Gait, Independent joint control, Co-contraction, Neuromechanics

Word count: 4568

Abstract

Background: Muscle co-activation, the simultaneous activation of muscle groups, is a common strategy to stabilize walking. However, co-activation can also be the consequence of underlying neurological impairments. This complicates differentiation between functional and pathological co-activation during walking. To better understand and discern functional co-activation during walking, this study investigated the difference between quadriceps-hamstrings co-activation during the swing phase of walking and isolated leg-swinging in healthy adults.

Methods: Twelve healthy young adults performed walking and isolated leg-swinging at slow (0.6 m/s) and comfortable speed. Electromyography signals from m. vastus lateralis, m. rectus femoris, m. biceps femoris, and m. semitendinosus were recorded. Co-activation index (CI) was calculated using Pearson correlation coefficient and area under the curve (AUC) and averaged to one quadriceps-hamstrings CI per metric.

Results: The results showed a higher Pearson-CI during walking compared to isolated leg-swinging, specifically during mid- and terminal-swing at both speeds. AUC-CI, but not Pearson-CI was significantly different between the two speeds.

Conclusion: Quadriceps-hamstrings co-activation towards the end of the swing phase during walking reflects preparation for heel-strike, which is not present in isolated leg-swinging. Therefore, an isolated leg-swinging task could serve as a feasible method to distinguish pathological from functional muscle co-activation during walking.

1. Introduction

Moving around in our daily lives requires refined muscle control to adapt to task and environmental constraints. A common functional adaptation strategy is muscle co-activation, which, in this study, is defined as the simultaneous activation of antagonistic muscle groups. While an agonist muscle generates force according to the demanded joint torque, the co-activated antagonist counteracts by producing force in the opposite direction, resulting in stiffening of the joint (Latash, 2018). During able-bodied walking, muscle co-activation is adjusted to accommodate changes in walking speed (Akl et al., 2021), slope (Lay et al., 2007) and compliance of the surface (MacLellan & Patla, 2006). Furthermore, co-activation during walking increases with age (Lee et al., 2017; Piche et al., 2022) and following neuromuscular impairments (Mohammadyari Gharehbolagh et al., 2023; Rosa, Marques, Demain, & Metcalf, 2014). In patient populations, instead of an adaptive strategy, muscle co-activation can also be the direct effect of motor control deficits (Busse et al., 2006; Hortobágyi & Devita, 2006). Problematically, studies of muscle co-activation during walking alone cannot distinguish between these different sources of co-activation, limiting our interpretation and understanding.

To gain a comprehensive understanding of muscle co-activation during normal and pathological walking, the focus could be directed to the swing phase of walking. From a traditional perspective, it is described that the swinging leg behaves like a pendulum (Mochon & McMahon, 1980). As the swinging leg moves forward, it requires only minimal muscle activation to gradually convert the swinging leg's potential energy into kinetic energy (Kuo &

Donelan, 2010). In this view, activation of hip flexors (i.e., m. iliopsoas and m. rectus femoris) accelerates the leg into swing and towards the end, the leg is decelerated by activation of the hamstring muscles (Uchida & Delp, 2020). However, push-off requires quadriceps activation to stabilize the knee, while, at the same time, the hamstrings are active to flex the knee into swing and clear the toe from the ground (Goldberg et al., 2004; Sadeghi et al. 2002). In addition, towards the end of swing, activation of the hamstrings can be accompanied by activation of the antagonistic quadriceps to stiffen the knee joint as a mechanism for shock absorption at heel-strike (Strazza et al., 2017). Thus, the minimal muscle activation required to swing the leg forward co-occurs with muscle activation to meet the task constraints, in terms of stabilization and ground clearance, of walking.

An isolated leg-swinging task ('leg-swinging') that replicates the swing phase of walking, could be used as a model to differentiate between pathological and functional co-activation during walking in people with motor impairments. In this study, we test this model in able-bodied individuals. The main objective is to investigate whether quadriceps-hamstrings co-activation is higher during walking compared to isolated leg-swinging. Co-activation during the swing phase of walking is influenced by the functional constraints of push-off and heel-strike whilst isolated leg-swinging requires selective and independent activation of quadriceps and hamstrings. Therefore, we hypothesize that the co-activation during the swing phase of walking will be higher compared to isolated leg-swinging. The second objective is to assess in which part of the swing phase these differences would become evident. We hypothesize that towards the end of the swing phase of walking, co-activation will be the highest as muscle activation is generated to prepare the leg for weight acceptance. Furthermore, we anticipate that co-activation during walking is dependent on speed, and pathological populations walk at slower speeds than able-bodied individuals. Therefore, the third aim is to investigate the effect of speed on the differences in quadriceps-hamstrings co-activation in the swing movement of walking and a leg-swinging movement.

2. Methods

2.1 Participants

Twelve healthy young adults (7 males, 5 females, age: 22.3 ±1.8 years, body height: 1.77 ±0.08 m, body weight: 72.7 ±10 kg, dominant leg: 1/11 (L/R), leg length: 0.97 ±0.04 m) volunteered to participate in this study. The inclusion criterion was age between 18 and 25 years. Exclusion criteria were (1) inability to understand the study instruction in Dutch, (2) indications of orthopedic, neurological, cardiorespiratory, and behavioral that may affect walking, and (3) contra-indications for physical activity assessed by the Physical Activity Readiness Questionnaire (PARQ). The study procedures were approved by the medical ethical committee of the University Medical Center Groningen (NL83016.042.22) and in line with the Declaration of Helsinki (World Medical Association, 2013). Participants signed written informed consent before participation.

2.2 Experimental protocol

Participants visited the lab on a single occasion. Before the experimental trials, height, weight, and leg length were measured and participants were asked to indicate their sex and age. The

study protocol consisted of four experimental conditions, two walk conditions, and two swing conditions (Fig. 1) at slow walking speed (SWS) and comfortable walking speed (CWS). Slow walking speed at 0.6 m/s normalized to leg length (Hof, 1996) was imposed to replicate a pathological walking speed.

Participants walked on the treadmill in a six-minute familiarization trial during which their CWS was established. Participants were asked to indicate their CWS while the experimenter increased the belt speed gradually. The belt speed was then set to 0.3 m/s above the indicated CWS while the experimenter gradually decreased the belt speed, participants were again asked to indicate their CWS. The CWS used in the experiment was the highest indicated speed in the two trials.

In the walking conditions participants were asked to walk for four minutes on the treadmill. Participants were not allowed to touch the handrails and were fitted with a safety harness to prevent falls, without providing support. During the last minute of the walking condition, the mean swing frequency was calculated from the mean time from toe-off to heel-strike, and the corresponding mean swing distance was calculated. In the subsequent swing conditions participants had to swing their dominant leg at the calculated swing frequency indicated by a metronome. Visual feedback on the swing distance was provided by a projection on the screen in front of the treadmill.

The swing condition consisted of four 50-second swing periods with 10-second rest periods in between to prevent fatigue. Participants stood in an upright stance position and were allowed to lean slightly to the side for clearance. Participants were instructed to swing their leg without touching the treadmill surface while keeping the metronome pace. They were allowed to rest their arm, contralateral to the swinging leg, on the handrail that was adjusted to their elbow height. Participants wore a safety harness during all conditions and received a rest period of minimum three minutes in between experimental conditions. The sequence of the two paired experimental conditions (SWS walking – SWS leg-swinging and CWS walking – CWS leg swinging) was randomized between participants.

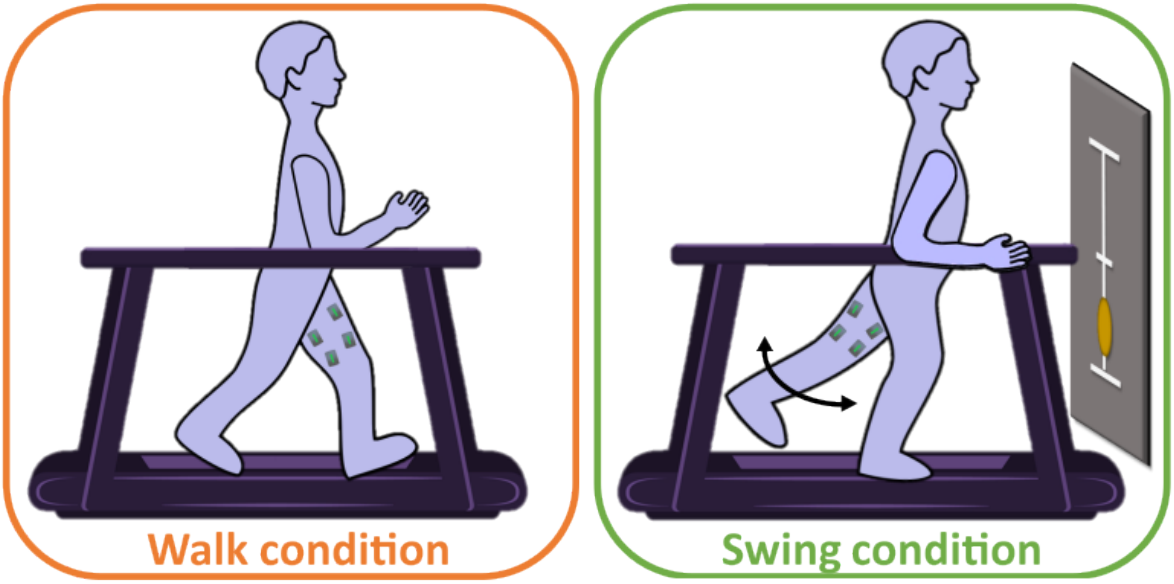

Fig 1. Illustration of the experimental set-up. In the swing condition, visual feedback on the amplitude of the swing was provided by a screen in front of the treadmill. During both conditions, electromyography signals of quadriceps and hamstrings,

kinematic and kinetic data were recorded. Motion capture cameras and passive markers on trochanter major and lateral malleolus are not included in the figure.

2.3 Instrumentation & data collection

Electromyography (EMG) data were recorded using surface electrodes (Delsys 16-channel sEMG system, Natick, MA, USA) at a sampling frequency of 2148 Hz. Electrodes were placed on four lower extremity muscles of the dominant leg (hamstrings: m. biceps femoris (BF), m. semitendinosus (ST); quadriceps: m. rectus femoris (RF), m. vastus lateralis (VL)). After the skin surface was shaved and cleaned with alcohol, the electrodes were placed according to the SENIAM conventions (Hermens et al., 1999). A three-dimensional motion capture system (10 cameras; Vicon Motion Systems Ltd, Yarnton, UK) recorded the trajectory of the ankle lateral malleolus and femur trochanter major of the dominant leg at a sampling frequency of 100 Hz. Three-dimensional ground reaction force (GRF, N), Center of pressure (COP, m), and three-dimensional moment (Nm) were measured with two force plates embedded in the treadmill (Motek Medical, Amsterdam, the Netherlands) at a sampling frequency of 1000 Hz. EMG, kinematic, and kinetic data were time-synchronized through a software trigger.

2.4 Data analysis

All data and statistical analyses were performed in MATLAB (version 2023b; The MathWorks Inc. Natick, MA, USA). The raw EMG signals were bandpass filtered with a 2nd order Butterworth filter (20-500 Hz) (Hermens et al., 1999) and full wave rectified. The signals were visually checked for artifacts. EMG amplitude normalization was done with respect to peak amplitude over all conditions (Besomi et al., 2020). After normalization, the first and last five seconds of each signal were removed, to exclude co-activation during initiation or and ending of walking and leg-swinging.

The COP data were high-pass filtered (5 Hz, 2nd order Butterworth filter) and low-pass filtered (10 Hz, 2nd order Butterworth filter) (Roerdink et al., 2008). The first and last five seconds of the COP signal were removed. The peaks in the anterior-posterior COP signal were used to detect foot contact events. The swing phase in the walking condition was defined as toe-off (TO) to heel-strike (HS) of the dominant leg. EMG data in the walking conditions were resampled from TO to ipsilateral HS on a 100-point time base.

For each of the four swing instances during walking and leg-swinging, the first and last five seconds were removed. The swing phase was defined as the maximal posterior position of the ankle to its maximal anterior position. Within this interval, EMG data were resampled for each swing instance on a 100-point time base. The four resampled EMG signals were averaged to one single swing EMG signal for each muscle.

A co-activation index (CI) was calculated for the quadriceps-hamstrings pairs (RF-BF, RF-ST, VL-BF and VL-ST) using Pearson correlation coefficient (Pearson-CI) (Field, 2016) and area under the curve (AUC; Eq. 1). The four quadriceps-hamstrings Pearson-CIs and AUC- CIs were averaged to one QD-HS Pearson-CI and one QD-HS AUC-CI for each time interval of interest in each condition. Positive Pearson-CI indicated co-activation, while negative Pearson-CI indicated no co-activation, i.e. the two muscles oppose each other in activation (Fig. 2).

$$\sum_{i=start}^{end}\left(\frac{\min\left(EMG_{muscle\ 1}(i), EMG_{muscle\ 2}(i)\right)}{\max\left(EMG_{muscle\ 1}(i), EMG_{muscle\ 2}(i)\right)}\right) * \frac{100\ \%}{\#\ of\ frames} \quad (1)$$

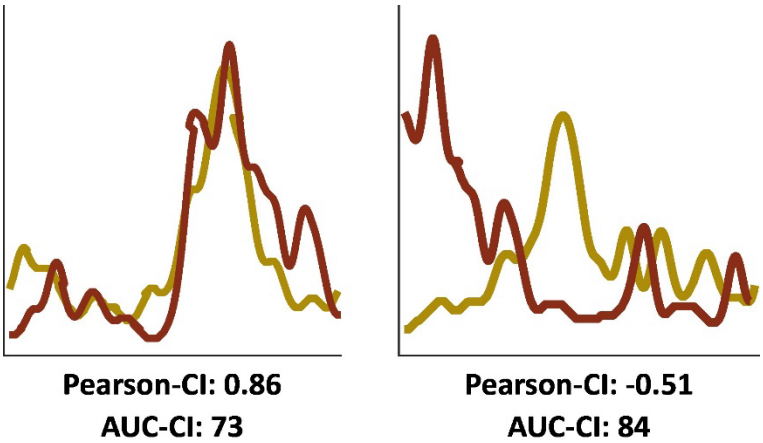

Fig. 2. Example of Pearson and area under the curve (AUC) co-activation indices (CI). Note that muscle activation and indices in the figure serve as examples and do not represent real data.

2.5 Statistical analysis

To assess whether swing time during leg-swinging matched swing time during walking we performed two paired t-tests to compare swing time between walking and leg-swinging for SWS and CWS separately.

Differences in co-activation of QD-HS between the walking and leg-swinging conditions were tested using two (Pearson-CI, AUC-CI) two-way (2*2) repeated measures ANOVAs with condition and speed as within-subjects factors. In case of violation of sphericity, Greenhouse Geiser corrected p-values were interpreted. Significant interaction effects were evaluated with Bonferroni post hoc corrections. Eta-squared (η^2) was reported and interpreted as 0.01 small effect size, 0.06 medium effect size 0.14 large effect size (Adams & Conway, 2014).

To investigate in which part of the swing phase differences in co-activation between walking and leg-swinging occurred, we first calculated a moving average of Pearson-CI and AUC-CI, using time bins each representing 25% of the swing phase. Then, four Statistical Parametric Mapping (SPM) paired t-tests (for each condition and each speed) were performed on the moving average signals. SPM is a technique that allows for the statistical analysis of temporal signals (Pataky et al., 2016). Statistical significance was set at $p < 0.05$ for all tests.

3. Results

The mean comfortable walking speed was 1.21 ±0.17 m/s and slow walking speed was fixed at 0.6 m/s normalized to leg length. SWS walking swing time (0.46 ±0.09 s) was not significantly different from SWS leg-swinging swing time (0.52 ±0.06 s, $t(1,11)=-1.821$, $p=0.096$).

Furthermore, CWS walking swing time (0.39 ±0.02 s) was not significantly different from CWS leg-swinging swing time (0.41 ±0.04 s, t(1,11)=-1.556, p=0.148).

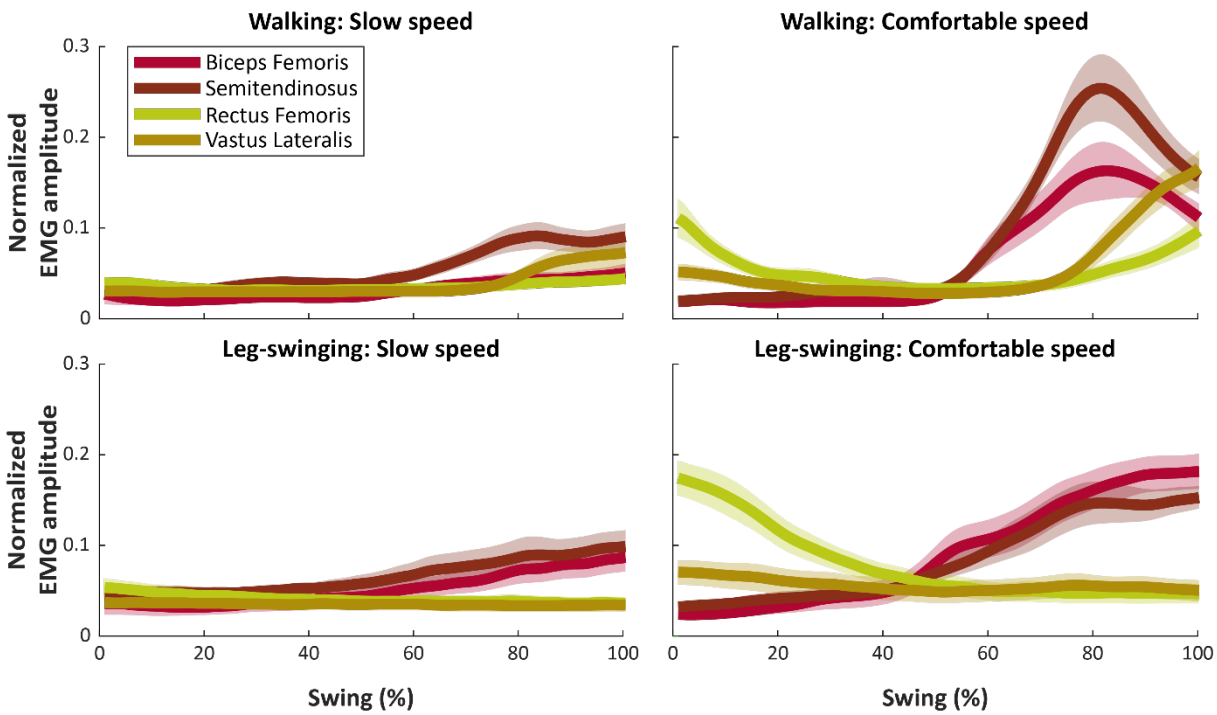

Fig. 3. Time normalized group-averaged muscle activity. Shaded areas around the mean represent the standard error of the mean.

3.1 Quadriceps-hamstrings co-activation during the full swing phase

Quadriceps and hamstrings muscle activation differed between condition and speed (Fig. 3). Co-activation was significantly higher in the swing phase of walking compared to isolated leg-swinging for Pearson-CI (Fig. 4; F(1,11)=62.131, p<0. 001, η^2=0.642), but not AUC-CI (F(1,11)=6.235, p=0.0297, η^2=0.035). The main effect of speed was significant for AUC-CI (F(1,11)=31.296, p<0.001, η^2=0.252), but not Pearson-CI (F(1,11)=2.450, p=0.146, η^2=0.012). Finally, no significant condition*speed interactions were found for Pearson-CI (F(1,11)=0.868, p=0.371, η^2=0.004) or AUC-CI (F(1,11)=2.215, p=0.164, η^2=0.016).

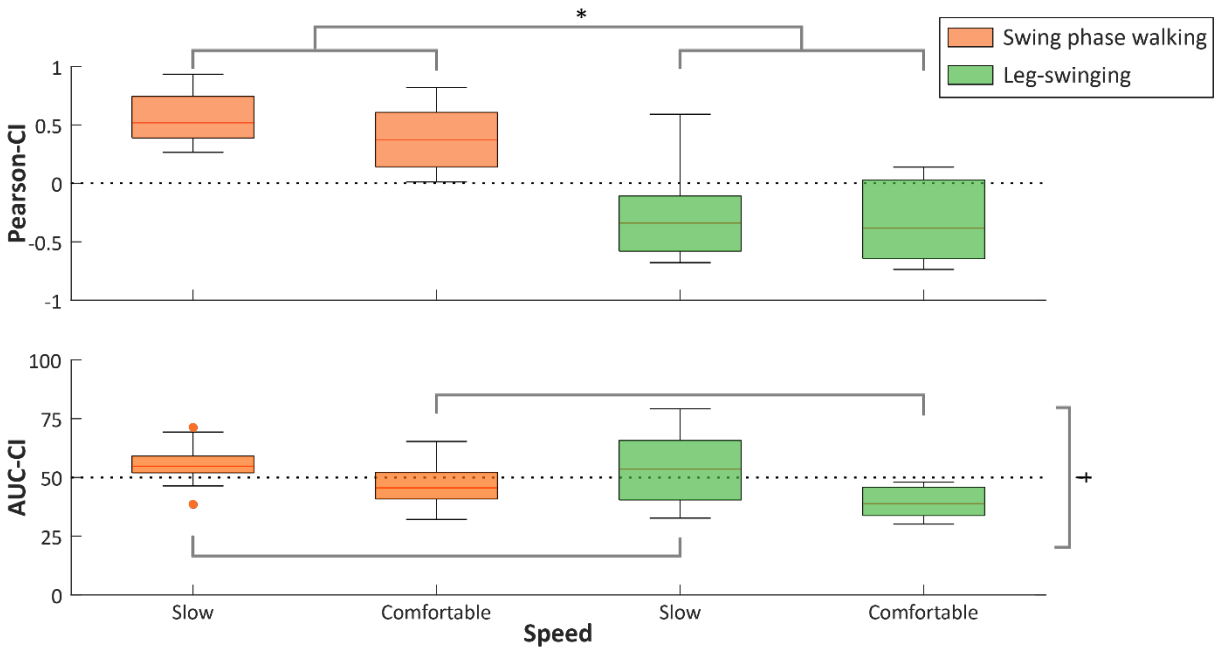

Fig. 4. Pearson and area under the curve AUC co-activation indices (CI) for slow walking speed (SWS; diamond-patterned boxes) and comfortable walking speed (CWS; solid boxes). * Indicates significant main effect for condition, † indicates significant main effect for speed.

3.2 Differences in quadriceps-hamstrings co-activation throughout the swing phase

SPM (Fig. 5) showed significantly higher Pearson-CI in SWS walking than SWS leg-swinging for windows 69-81 ($t(1,11)=3.380$, $p<0.001$). Furthermore, Pearson-CI was significantly higher in CWS walking than CWS leg-swinging for windows 52-57, windows 59-60, and windows 55-74 ($t(1,11)=3.454$, $p=0.015$, $p=0.048$, $p<0.001$ respectively). Finally, AUC-CI was significantly higher during CWS walking than CWS leg-swinging for windows 85-87 ($t(1,11)=2.974$, $p=0.048$).

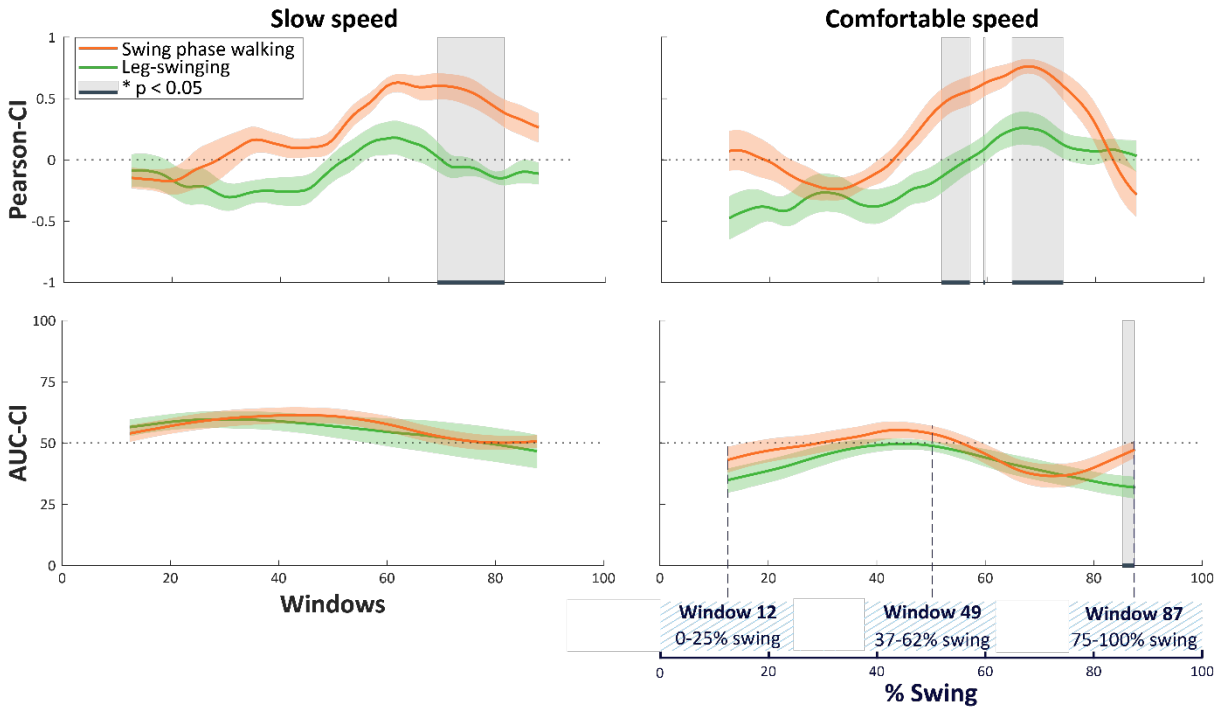

Fig. 5. SPM results showing group averaged Pearson and area under the curve (AUC) co-activation indices (CI) for each window representing 25% of the swing condition. Shaded areas around the mean represent the standard error of the mean. The lower-right panel indicates how three windows represent percentages of the original swing phase.

4. Discussion

This study aimed to investigate whether quadriceps-hamstrings co-activation is higher during walking compared to isolated leg-swinging. The results showed higher Pearson-CI during walking compared to leg-swinging but no differences in AUC-CI. Specifically, Pearson-CI was higher during walking than isolated leg-swinging in mid- and terminal-swing. Furthermore, the results showed no effect of speed on Pearson-CI but did show an effect of speed on AUC-CI. These results indicate that quadriceps-hamstrings co-activation is related to the task constraints of walking and therefore higher compared to isolated leg-swinging. This is in line with previous studies, which demonstrated the functional co-activation of quadriceps and hamstrings during abled-bodied walking based on the full walking cycle (Akl et al., 2021; Mengarelli et al., 2018; Strazza et al., 2017).

Our results indicate that quadriceps-hamstrings co-activation is not different between walking and leg-swinging during the initial part of the swing phase, but that differences occur during mid and terminal swing. In the swing phase of walking, the m. rectus femoris (and m. iliopsoas) accelerate the leg forward (Uchida & Delp, 2020). Following mid-swing, the eccentric contraction of the hamstrings absorbs the kinetic energy of the forward moving leg, slowing down its forward movement (Ivanenko et al., 2004; Neptune et al., 2009). The quadriceps remain activated and quadriceps-hamstrings co-activation stabilize the knee joint in preparation for weight acceptance (Strazza et al., 2017; Neptune et al., 2009). This rather efficient strategy in able-bodied walking, which is reflected by co-activation, applies to a lesser extent to neurological populations, in which muscle co-activation during walking is shown to be more pronounced due to underlying neurological impairments, resulting in a loss of independent joint control (Mari et al., 2014; Mohammadyari Gharehbolagh et al., 2023; Rosa,

Marques, Demain, & Metcalf, 2014). In these clinical populations, co-activation during the leg-swinging task would indicate co-activation inherent to their underlying neuropathy rather than a strategy to meet the task constraints. Able-bodied individuals can selectively control their hamstrings and quadriceps muscles. However, our results show that this cannot be inferred from muscle activation during the swing phase of walking alone. The current isolated leg-swinging approach allows to distinguish muscle co-activation caused by underlying neurological impairments from task-related muscle co-activation during walking. Moreover, when able-bodied young adults were asked to swing their leg at speeds similar to pathological walking speeds, they did not exhibit differences in co-activation. As such, the leg-swinging task could be used to assess co-activation due to muscle coordination deficits during walking in clinical populations in future research.

The current results show no significant condition*speed interactions, which indicates that, although the co-activation timing was different, the phasing of the co-activation between conditions is comparable. Several studies have shown consistencies in muscle activation phasing during walking across different walking speeds (Buurke et al., 2016; Den Otter et al., 2004; Kibushi et al., 2018). In the current study, AUC-CI was different between SWS and CWS. An explanation for this finding can be the ongoing diminished muscle activation during SWS conditions compared to CWS conditions, i.e. as an effect of speed EMG amplitude is lower at slower walking speeds (Den Otter et al., 2004). As such, the maximum amplitude is lower in SWS than in CWS, which may inflate the AUC-CI.

The disparities between Pearson-CI and AUC-CI demonstrate that whether muscles are classified to be co-activated or not is highly dependent on the methodological decision for the co-activation metric. Pearson-CI indicates whether the magnitude of the activation of a muscle is associated with the change in the magnitude of the activation of another muscle. This association is either in the same direction (positive CI: co-activation) or in the opposite direction (negative CI: no co-activation) (Schober & Schwarte, 2018). This metric is robust against any amplitude normalization method, but unlike other methods (Rosa, et al., 2014; Souissi et al., 2017) does not take the timing and magnitude of the co-activation into account. One could argue that two muscles co-activated in the same direction but with only a small magnitude of short-to-modest duration only contribute to a small part of the total force production (Staudenmann et al., 2010). To indicate the relative magnitude of the co-activation, we expressed co-activation in an additional AUC-CI. However, as mentioned before, AUC may become inflated in instances in which both muscles are of relatively low amplitude. Currently, the use of the co-activation metric lacks consensus and the results of this study further emphasize a clear and cautious definition of the co-activation metric in future studies.

A methodological consideration was the use of averaged quadriceps co-activation from four distinct individual QD-HS muscle pairs. m. biceps femoris, m. semitendinosus, and m. rectus femoris are bi-articular muscles and act on the knee and hip joint, while the m. vastus lateralis only spans the knee joint. This approach could potentially oversimplify the complexities of upper limb muscle activation during swing. However, this consideration increases this study's statistical power as the focus was the difference in general quadriceps-hamstrings muscle co-activation during walking and isolated leg-swinging.

It can be concluded that quadriceps-hamstrings co-activation towards the end of the swing phase of walking reflects simultaneous deceleration of the leg and joint-stiffening in

preparation for heel-strike, because this co-activation is not presented in isolated leg-swing. The leg-swinging task presented here could serve as a feasible method to distinguish pathological muscle co-activation from task-related muscle co-activation during walking in the future. The degree of co-activation is greatly influenced by the methodological considerations for the co-activation metric, which emphasizes the need for caution when interpreting co-activation indices and the need for consensus the definition and interpretation of co-activation metrics.

Acknowledgements

The authors wish to thank Anniek Heerschop, Emyl Smid, Dirk van der Meer, Wim Kaan, Daniel Dias dos Santos, David Hozeman, Isa Kreuwel and Marije Bouius for their help.

Conflict of interest

The authors declare no conflicts of interest.

Funding

Tom Buurke was funded by Research Foundation - Flanders (FWO): 12ZJ922N. The sponsor had no involvement in the design, data collection, or writing of the manuscript.

References

Adams, M. A., & Conway, T. L. (2014). Eta Squared. In A. C. Michalos (Ed.), *Encyclopedia of Quality of Life and Well-Being Research* (pp. 1965–1966). Springer Netherlands.

Akl, A. R., Gonçalves, P., Fonseca, P., Hassan, A., Vilas-Boas, J. P., & Conceição, F. (2021). Muscle co-activation around the knee during different walking speeds in healthy females. *Sensors (Switzerland)*, *21*(3), 1–12.

Besomi, M., Hodges, P. W., Clancy, E. A., Van Dieën, J., Hug, F., Lowery, M., Merletti, R., Søgaard, K., Wrigley, T., Besier, T., Carson, R. G., Disselhorst-Klug, C., Enoka, R. M., Falla, D., Farina, D., Gandevia, S., Holobar, A., Kiernan, M. C., McGill, K., … Tucker, K. (2020). Consensus for experimental design in electromyography (CEDE) project: Amplitude normalization matrix. *Journal of Electromyography and Kinesiology*, *53*(June), 102438.

Busse, M. E., Wiles, C. M., & Van Deursen, R. W. M. (2006). Co-activation: Its association with weakness and specific neurological pathology. *Journal of NeuroEngineering and Rehabilitation*, *3*.

Buurke, T. J. W., Lamoth, C. J. C., Van Woude, L. H. V. D., & Otter, A. R. Den. (2016). Synergistic structure in the speed dependent modulation of muscle activity in human walking. *PLoS ONE*, *11*(4), 1–19.

Crenna, F., Rossi, G. B., & Berardengo, M. (2021). Filtering biomechanical signals in movement analysis. *Sensors*, *21*(13).

Den Otter, A. R., Geurts, A. C. H., Mulder, T., & Duysens, J. (2004). Speed related changes in muscle activity from normal to very slow walking speeds. *Gait and Posture*, *19*(3), 270–278.

Field, A. (2016). *Discovering Statistics Using IBM SPSS Statistics* (4th ed.). SAGE Publications.

Goldberg, S. R., Anderson, F. C., Pandy, M. G., & Delp, S. L. (2004). Muscles that influence knee flexion velocity in double support: Implications for stiff-knee gait. *Journal of Biomechanics*, *37*(8), 1189–1196.

Hermens, H. J., Freriks, B., Merletti, R., Stegeman, D., Blok, J., Rau, G., Disselhorst-Klug, C., & Hägg, G. (1999). European Recommendations for Surface ElectroMyoGraphy. *Roessingh Research and Development*, 8–11.

Hof, A.L. (1996) Scaling gait data to body size. *Gait & Posture, 4*, 222-223.

Hortobágyi, T., & Devita, P. (2006). Mechanisms Responsible for the Age-Associated Increase in Coactivation of Antagonist Muscles. In *Exerc. Sport Sci. Rev* (Vol. 34, Issue 1).

Ivanenko, Y. P., Poppele, R. E., & Lacquaniti, F. (2004). Five basic muscle activation patterns account for muscle activity during human locomotion. *Journal of Physiology*, *556*(1), 267–282.

Kibushi, B., Hagio, S., Moritani, T., & Kouzaki, M. (2018). Speed-dependent modulation of muscle activity based on muscle synergies during treadmill walking. *Frontiers in Human Neuroscience*, *12*(January), 1–13.

Kuo, A. D., & Donelan, J. M. (2010). Dynamic Principles of Gait and Their Clinical Implications. In *Physical Therapy* (Vol. 90, Issue 2).

Latash, M. L. (2018). Muscle coactivation: Definitions, mechanisms, and functions. *Journal of Neurophysiology*, *120*(1), 88–104.

Lay, A. N., Hass, C. J., Richard Nichols, T., & Gregor, R. J. (2007). The effects of sloped surfaces on locomotion: An electromyographic analysis. *Journal of Biomechanics*, *40*(6), 1276–1285.

Lee, H. J., Chang, W. H., Choi, B. O., Ryu, G. H., & Kim, Y. H. (2017). Age-related differences in muscle co-activation during locomotion and their relationship with gait speed: a pilot study. *BMC Geriatrics*, *17*(1), 1–8. https://doi.org/10.1186/s12877-017-0417-4

MacLellan, M. J., & Patla, A. E. (2006). Adaptations of walking pattern on a compliant surface to regulate dynamic stability. *Experimental Brain Research*, *173*(3), 521–530.

Mari, S., Serrao, M., Casali, C., Conte, C., Martino, G., Ranavolo, A., Coppola, G., Draicchio, F., Padua, L., Sandrini, G., & Pierelli, F. (2014). Lower limb antagonist muscle co-activation and its relationship with gait parameters in cerebellar ataxia. *Cerebellum*, *13*(2), 226–236.

World Medical Association. World Medical Association Declaration of Helsinki: ethical principles for medical research involving human subjects. JAMA. 2013 Nov 27;310(20):2191-4. doi: 10.1001/jama.2013.281053. PMID: 24141714

Mengarelli, A., Gentili, A., Strazza, A., Burattini, L., Fioretti, S., & Di Nardo, F. (2018). Co-activation patterns of gastrocnemius and quadriceps femoris in controlling the knee joint during walking. *Journal of Electromyography and Kinesiology*, *42*, 117–122.

Mochon, S., & Mcmahon, T. A. (1980). *BALLISTIC WALKING** (Vol. 13).

Mohammadyari Gharehbolagh, S., Dussault-Picard, C., Arvisais, D., & Dixon, P. C. (2023). Muscle co-contraction and co-activation in cerebral palsy during gait: A scoping review. In *Gait and Posture* (Vol. 105, pp. 6–16). Elsevier B.V.

Roerdink, M., Coolen, B. H., Clairbois, B. H., Lamoth, C. J., & Beek, P. J. (2008). Online gait event detection using a large force platform embedded in a treadmill. *Journal of biomechanics*, *41*(12), 2628–2632.

Neptune, R. R., Clark, D. J., & Kautz, S. A. (2009). Modular control of human walking: A simulation study. *Journal of Biomechanics*, *42*(9), 1282–1287.

Pataky, T. C., Robinson, M. A., & Vanrenterghem, J. (2016). Region-of-interest analyses of onedimensional biomechanical trajectories: Bridging 0D and 1D theory, augmenting statistical power. *PeerJ*, *2016*(11).

Piche, E., Chorin, F., Zory, R., Duarte Freitas, P., Guerin, O., & Gerus, P. (2022). Metabolic cost and co-contraction during walking at different speeds in young and old adults. *Walking and Posture*, *91*(October 2021), 111–116.

Review, C., Communication, S., & Principles, G. (2014). World Medical Association Declaration of Helsinki: ethical principles for medical research involving human subjects. *The Journal of the American College of Dentists*, *81*(3), 14–18.

Rosa, M. C. N., Marques, A., Demain, S., & Metcalf, C. D. (2014). Lower limb co-contraction during walking in subjects with stroke: A systematic review. *Journal of Electromyography and Kinesiology*, *24*(1), 1–10.

Rosa, M. C. N., Marques, A., Demain, S., Metcalf, C. D., & Rodrigues, J. (2014). Methodologies to assess muscle co-contraction during gait in people with neurological impairment - A systematic literature review. *Journal of Electromyography and Kinesiology*, *24*(2), 179–191.

Sadeghi H, Allard P, Barbier F, Sadeghi S, Hinse S, Perrault R, Labelle H. Main functional roles of knee flexors/extensors in able-bodied gait using principal component analysis (I). Knee. 2002 Feb;9(1):47-53.

Schober, P., & Schwarte, L. A. (2018). Correlation coefficients: Appropriate use and interpretation. *Anesthesia and Analgesia*, *126*(5), 1763–1768.

Souissi, H., Zory, R., Bredin, J., & Gerus, P. (2017). Comparison of methodologies to assess muscle co-contraction during gait. *Journal of Biomechanics*, *57*, 141–145.

Staudenmann, D., Roeleveld, K., Stegeman, D. F., & van Dieen, J. H. (2010). Methodological aspects of SEMG recordings for force estimation - A tutorial and review. In *Journal of Electromyography and Kinesiology* (Vol. 20, Issue 3, pp. 375–387).

Strazza, A., Mengarelli, A., Fioretti, S., Burattini, L., Agostini, V., Knaflitz, M., & Di Nardo, F. (2017). Surface-EMG analysis for the quantification of thigh muscle dynamic co-contractions during normal walking. *Walking and Posture*, *51*, 228–233.

Uchida, T., & Delp, S. (2020). Biomechanics of Movement: The Science of Sports, Robotics, and Rehabilitation. The MIT Press.