2021 43rd Annual International Conference of the
IEEE Engineering in Medicine & Biology Society (EMBC)
Oct 31 - Nov 4, 2021. Virtual Conference

Lower leg muscle force prediction in gait transition

Camila Taira, Masayuki Kawada, Ryoji Kiyama, Arturo Forner-Cordero, *Senior Member, IEEE*

Abstract— **Walking and running, the two most basic and functional gait modes, have been often addressed through EMG, kinematics and biomechanical modelling, however, there is no consensus in the literature on which factors trigger the transition from walking to running. Ankle plantarflexors and dorsiflexor were found to play an important role in gait transition due to higher muscular activation to propel the body forward to run. We tested these muscles activation during walking and running at the same speeds, through a musculoskeletal model derived from subjects' kinematic and kinetic data. Compared to EMG data frequently reported in the literature, the results yielded similar activation patterns for all muscles analyzed. Besides, across speeds, dorsiflexor activation kept increasing in walking, especially after PTS (preferred transition speed), which may indicate its contribution to gait transition, as an effort to bring the foot forward to keep up with the unnatural condition of walking at high speeds.**

I. INTRODUCTION

Humans change locomotion mode depending on the speed. Several factors have been reported to be associated with the walk-to-run transition (WRT). Walking and running, as the most common and functional locomotion modes, have been extensively studied and their differences reported in the literature [1]–[4]. Nevertheless, the walk-to-run transition remains not well-understood.

Minimization of metabolic cost alone cannot fully explain the WRT [5]. Perceived effort of lower limb muscles during walking is also related to the WRT. Ankle plantarflexors and dorsiflexors presented the most pronounced muscular change during the WRT [6]. While some studies pointed to dorsiflexors as determinant to gait transition [5], [7], [8] others found plantarflexors to highly contribute to the switch from walking to running [9], [10].

Some studies found these muscles EMG and/or simulated activation and/or joint angles and moments varying along with speeds in walking and running [3], [5]–[7] , but they did not analyze their activation in both conditions at the same speeds (percentages of each subjects' preferred transition speed (PTS)) through a musculoskeletal model.

Biomechanical modelling and simulations to estimate muscle force have been widely used in gait research over the last decades [11]–[19] and often comparison of simulation data from musculoskeletal models with EMG experimental data is used to corroborate or validate methods and results [4], [9], [15], [20]–[23] or to predict muscle behavior in pathological or extreme and unusual conditions [24]–[27].

Therefore, the aim of this study was to investigate the contribution of the lower leg muscles in the WRT. In order to do so, we analyzed gastrocnemius (GAS), soleus (SOL) and tibialis anterior (TA) activation during walking and running at the same speeds (around each subjects' transition speed) through a musculoskeletal model based on their own kinematic and kinetic data.

II. METHODS

A. Subjects

Kinematic data and ground reaction force from 10 healthy subjects (age 26± 5.4 years old; height 169.1± 5.6; weight 63.7± 9.9) were recorded through VICON (MX3, Oxford Metrics) and two force plates (OR-6-7-2000, BP400600-2000 Advanced Mechanical Technology, Inc., Model, Watertown, MA, USA). The study was approved by the Clinical Research Review Board of the Faculty of Medicine, Kagoshima University (N. 205). All subjects signed an informed consent.

To determine their walk-to-run transition speed (WRT-speed), they started walking on a treadmill, at 1.25 m/s and the speed was increased 0.1 m/s every 30 seconds, until they started running, and this speed was considered their WRT-speed.

B. Experimental procedure

Kinematic and kinetic data from the subjects were used as input to a musculoskeletal model to simulate muscle activation of lower leg muscles during walking and running at the same speeds

Experimental kinematic data recording

Infrared reflective markers of 14mm diameter were attached to the following skin landmarks: right and left, front and back head; C1; clavicle; sternum; T10; right and left, anterior and posterior superior iliac spine; left trocanter, thigh, knee, ankle, heel, first and fifth metatarsophalangeal.

Subjects were asked to run and walk gradually faster, and data matched to 60%, 80%, 100%, 120% and 140% of their PTS were used for later analysis. The data were taken for the left leg and after that 10 gait or running cycles were chosen.

We calculated muscle force from the trajectory of those

*This research was supported by MEXT Japan.

C. Taira and A. Forner-Cordero are from the Biomechatronics Laboratory. Mechatronics Dept. Escola Politécnica, University of São Paulo, Brazil. E-mail: catai@usp.br, aforner@usp.br

M. Kawada and R. Kiyama are from the Biomechanics Laboratory, Dept. of Physical Therapy. School of Health Sciences, Faculty of Medicine, Kagoshima University, Japan. E-mail: kiyama@health.nop.kagoshima-u.ac.jp

978-1-7281-1178-0/21/$31.00 ©2021 IEEE

markers attached on anatomical landmarks, recorded by a Vicon motion capture system, and ground reaction forces recorded by a force platform.

Musculoskeletal Model

A musculoskeletal simulation software AnyBody 7.3.1 (Anybody Technology, Aalborg, Denmark) was used to calculate activation of the lower leg muscles from subjects' kinematic and kinetic data. The musculoskeletal model consisted of 5 rigid bodies: trunk, pelvis, femur, tibia, foot, driven by 170 actuators representing 39 Hill-type muscles. Muscle force was calculated using optimization algorithm to minimize the 3rd power of all muscles' stress. Simulated muscle activation was the ratio of the calculated muscle force with respect to its maximum capacity.

We analyzed the left leg considering that walking and running are symmetrical movements. We previously confirmed the validity of muscle force from the 5-segment model. Among muscle force estimation of all muscles of left lower limb, three main muscles were considered and reported here: gastrocnemius (GAS), soleus (SOL) and tibialis anterior (TA) to calculate the ankle joint torque.

C. Data Analysis

Time of each trial was normalized to percentage of the gait or running cycle. A total of 10 gait or running cycles were ensemble averaged for further analysis. The averaged kinematic and kinetic data were used to estimate muscle activation. Then, peak value was calculated from simulated muscle activation, and normalized by the maximum of 100% PTS walking. Two-way repeated measures ANOVA was applied to test for significant differences between the five speeds and the two locomotion modes (walking and running). If there was significant interaction, differences between conditions were tested by paired t-test for the simulated muscle activation. Statistics was performed using SPSS (Version 20.0 for Windows: Chicago, USA). It was considered significant a p-value of 0.05.

III. RESULTS

The average PTS was 1.83±0.12 m/s. The stride length showed interaction between the two conditions and velocities ($F_{4,36}$ = 122.11, P<0.001). and it was significantly greater in walking across all speeds, except at 140% PTS (Fig 1).

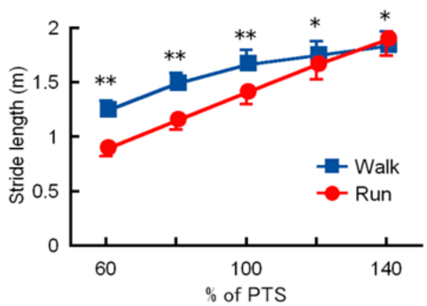

Fig 1. Averaged stride length during walking (blue) and running (red) at 60%, 80%, 100%, 120% and 140% of the PTS. An asterisk indicates statistically significant differences (P<0.05), and double asterisk indicates (P<0.01).

The figures present the results for each muscle in the following order: GAS (Figs 2 and 3), SOL (Figs 4 and 5) and TA (Figs 6 and 7). The first figure for each muscle represents activation waveforms throughout a whole gait cycle and the second figure of each muscle shows its activation peak values across speeds (%PTS).

The simulated activation patterns are similar to those experimentally measured EMG patterns from literature [2], [9], [21], [28].

Interaction between the five speeds and the two locomotion modes was observed in TA normalized simulated activation at the 1st (F = 13.19, P < 0.001) and 2nd peaks (F = 12.8, P < 0.001), and GAS (F = 8.00, P < 0.001).

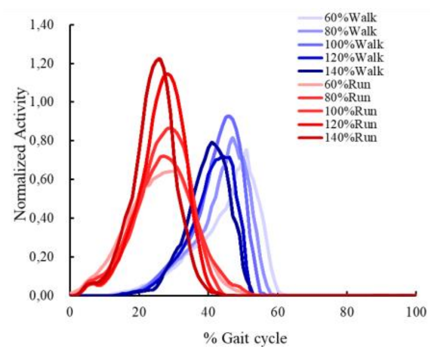

Fig 2. Gastrocnemius normalized simulated activation pattern in walking (blue) and running (red) across speeds.

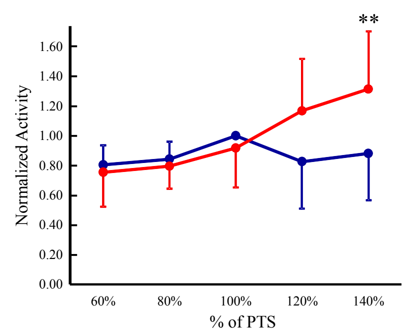

Fig 3. Gastrocnemius peak values of simulated activation in walking (blue) and running (red) across %PTS. Statistically significant differences are indicated either with a single asterisk (P<0.05), or double asterisk (P<0.01).

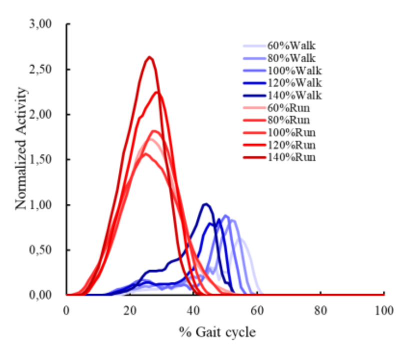

Fig 4. Soleus normalized simulated activation pattern in walking (blue) and running (red) across speeds.

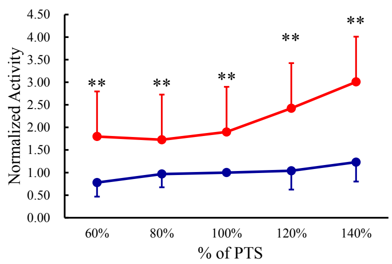

Fig 5. Soleus peak values of simulated activation in walking (blue) and running (red) across %PTS. Statistically significant differences are indicated either with a single asterisk (P<0.05), or double asterisk (P<0.01).

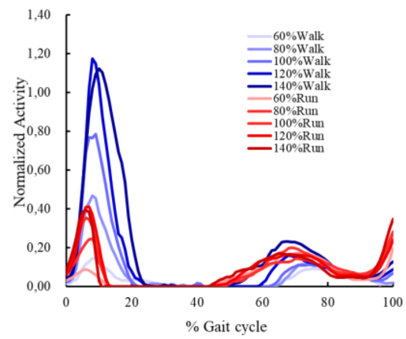

Fig 6. Tibialis Anterior normalized simulated activation pattern in walking (blue) and running (red) across speeds.

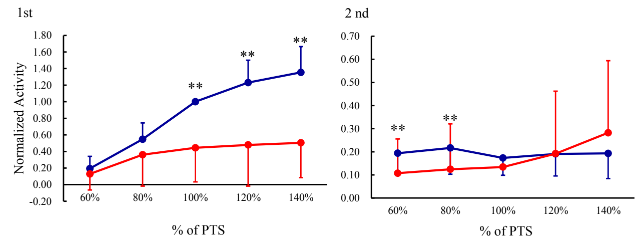

Fig 7. Tibialis Anterior first and second peak values of simulated activation in walking (blue) and running (red) across %PTS. Statistically significant differences are indicated either with a single asterisk (P<0.05), or a double one (P<0.01).

Contrary to TA, SOL showed significantly higher activation in running for all speeds (Fig 5). Whereas GAS only had higher activation in running for the two higher speeds (Fig 3).

TA activation at the 1st peak during walking was higher than running for all five speeds analyzed and significant differences were observed at 100 % PTS and above. For its 2nd peak, TA activation was significantly higher in walking at low speeds, before PTS. Only after 120% PTS, activation in running becomes higher (Fig 7).

IV. Discussion

We compared GAS, SOL and TA simulated muscle force in walking and running at the same five speeds 60%, 80%, 100%, 120% and 140% of the PTS.

There are plenty of studies on walking and running in the literature analyzing lower leg [1]–[3], [5], [6], [24]. However, there is a lack of research in the literature that analyzed and compared, through a musculoskeletal model, both gait modes throughout the whole gait cycle and at the same functional speeds around the PTS for the main lower leg muscles.

At TA 1st peak, there was a tendency to keep increasing its activation values in walking along with increase on speeds, agreeing with previous studies [22], [29], especially after the PTS [28]. This is possibly due to a higher dorsiflexion torque demand after heel-strike to keep walking at high speeds when it would be more natural to run.

TA 2nd peak, prior to swing phase, is necessary to avoid foot scuffing, preparing the foot for the next heel strike [31]. This peak comes after ankleplantarflexors burst, so to propel the ankle forward, dorsiflexors must overcome ankle plantarflexor contraction [31].

Based on such TA activation peak after PTS in walking, gait transition could be expected to avoid larger activation peaks after heel-strike (first peak).

Therefore, in agreement with previous research, increasingly high activation levels of TA after PTS indicate muscle stress and could trigger the transition to running. There are previous studies that found TA to be a determinant factor for gait transition [5], [28], [31].

GAS and SOL activation increased above 100% PTS in running due to larger GRF and forefoot contact with the ground. There was a marked difference in activation of both muscles between the two conditions, having their peaks in running earlier than in walking, which is a well-known difference between gait modes. However, their activation did not significantly change after PTS, neither peaked in a way that could indicate muscle overexertion.

Finally, there are different aspects that should be addressed in future research. For example, to include other biomechanical measurements, as kinematic data, such as arm and trunk motion, kinetic and ground reaction forces. Not only other musculoskeletal models should be tested but also, a wide range of conditions, natural and pathological, can also be predicted and studied by changing model parameters.

Acknowledgments

The authors would like to thank all the volunteers that participated in the experiments. CT thanks the Ministry of Education, Culture, Sports, Science and Technology - Japan (MEXT) for a Master Science grant, number 090613. AFC acknowledges a grant from CNPq 312236/2019-0.

References

[1] A. A. Biewener, C. T. Farley, T. J. Roberts, and M. Temaner, “Muscle mechanical advantage of human walking and running: Implications for energy cost,” *J. Appl. Physiol.*, vol. 97, no. 6, pp. 2266–2274, 2004, doi: 10.1152/japplphysiol.00003.2004.

[2] G. Cappellini, Y. P. Ivanenko, R. E. Poppele, and F. Lacquaniti, “Motor patterns in human walking and running,” *J. Neurophysiol.*, vol. 95, no. 6, pp. 3426–3437, 2006, doi: 10.1152/jn.00081.2006.

[3] M. Ishikawa, J. Pakaslahti, and P. V. Komi, "Medial gastrocnemius muscle behavior during human running and walking," *Gait Posture*, vol. 25, no. 3, pp. 380–384, 2007, doi: 10.1016/j.gaitpost.2006.05.002.
[4] K. Sasaki and R. R. Neptune, "Differences in muscle function during walking and running at the same speed," *J. Biomech.*, vol. 39, no. 11, pp. 2005–2013, 2006, doi: 10.1016/j.jbiomech.2005.06.019.
[5] V. Segers, M. Lenoir, P. Aerts, and D. De Clercq, "Influence of M. tibialis anterior fatigue on the walk-to-run and run-to-walk transition in non-steady state locomotion," *Gait Posture*, vol. 25, no. 4, pp. 639–647, 2007, doi: 10.1016/j.gaitpost.2006.07.008.
[6] J. Nilsson, A. Thorstensson, and J. Halbertsma, "Changes in leg movements and muscle activity with speed of locomotion and mode of progression in humans," *Acta Physiol. Scand.*, vol. 123, no. 4, pp. 457–475, 1985, doi: 10.1111/j.1748-1716.1985.tb07612.x.
[7] A. Hreljac, A. Arata, R. Ferber, J. A. Mercer, and B. S. Row, "An electromyographical analysis of the role of dorsiflexors on the gait transition during human locomotion," *J. Appl. Biomech.*, vol. 17, no. 4, pp. 287–296, 2001, doi: 10.1123/jab.17.4.287.
[8] N. J. Pires, B. S. Lay, and J. Rubenson, "Joint-level mechanics of the walk-to-run transition in humans," *J. Exp. Biol.*, vol. 217, no. 19, pp. 3519–3527, Oct. 2014, doi: 10.1242/jeb.107599.
[9] R. R. Neptune and K. Sasaki, "Ankle plantar flexor force production is an important determinant of the preferred walk-to-run transition speed.," *J. Exp. Biol.*, vol. 208, no. Pt 5, pp. 799–808, Mar. 2005, doi: 10.1242/jeb.01435.
[10] M. G. Pandy and T. P. Andriacchi, *Muscle and Joint Function in Human Locomotion*, vol. 12, no. 1. 2010.
[11] U. Trinler, H. Schwameder, R. Baker, and N. Alexander, "Muscle force estimation in clinical gait analysis using AnyBody and OpenSim," *J. Biomech.*, vol. 86, pp. 55–63, 2019, doi: 10.1016/j.jbiomech.2019.01.045.
[12] H. Kainz *et al.*, "Reliability of four models for clinical gait analysis," *Gait Posture*, vol. 54, no. November 2016, pp. 325–331, 2017, doi: 10.1016/j.gaitpost.2017.04.001.
[13] A. Leardini, C. Belvedere, F. Nardini, N. Sancisi, M. Conconi, and V. Parenti-Castelli, "Kinematic models of lower limb joints for musculo-skeletal modelling and optimization in gait analysis," *J. Biomech.*, vol. 62, pp. 77–86, 2017, doi: 10.1016/j.jbiomech.2017.04.029.
[14] L. Modenese, A. T. M. Phillips, and A. M. J. Bull, "An open source lower limb model: Hip joint validation," *J. Biomech.*, vol. 44, no. 12, pp. 2185–2193, 2011, doi: 10.1016/j.jbiomech.2011.06.019.
[15] A. S. Arnold, M. H. Schwartz, D. G. Thelen, and S. L. Delp, "Contributions of muscles to terminal-swing knee motions vary with walking speed," *J. Biomech.*, vol. 40, no. 16, pp. 3660–3671, 2007, doi: 10.1016/j.jbiomech.2007.06.006.
[16] M. Kawada, K. Hata, R. Kiyama, T. Maeda, and K. Yone, "Biomechanical characterization of slope walking using musculoskeletal model simulation," *Acta Bioeng. Biomech.*, vol. 20, no. 1, pp. 117–125, 2018, doi: 10.5277/ABB-01000-2017-01.
[17] M. G. Hoy, F. E. Zajac, and M. E. Gordon, "A musculoskeletal model of the human lower extremity: The effect of muscle, tendon, and moment arm on the moment-angle relationship of musculotendon actuators at the hip, knee, and ankle," *J. Biomech.*, vol. 23, no. 2, pp. 157–169, 1990, doi: 10.1016/0021-9290(90)90349-8.
[18] S. Siegler, R. Seliktar, and W. Hyman, "Simulation of human gait with the aid of a simple mechanical model," *J. Biomech.*, vol. 15, no. 6, pp. 415–425, 1982, doi: 10.1016/0021-9290(82)90078-1.
[19] M. Kawada *et al.*, "Contribution of hip and knee muscles to lateral knee stability during gait," *J. Phys. Ther. Sci.*, vol. 32, no. 11, pp. 729–734, 2020, doi: 10.1589/jpts.32.729.
[20] S. R. Hamner and S. L. Delp, "Muscle contributions to fore-aft and vertical body mass center accelerations over a range of running speeds," *J. Biomech.*, vol. 46, no. 4, pp. 780–787, Feb. 2013, doi: 10.1016/j.jbiomech.2012.11.024.
[21] R. R. Neptune, K. Sasaki, and S. A. Kautz, "The effect of walking speed on muscle function and mechanical energetics," *Gait Posture*, vol. 28, no. 1, pp. 135–143, 2008, doi: 10.1016/j.gaitpost.2007.11.004.
[22] M. Q. Liu, F. C. Anderson, M. H. Schwartz, and S. L. Delp, "Muscle contributions to support and progression over a range of walking speeds," *J. Biomech.*, vol. 41, no. 15, pp. 3243–3252, 2008, doi: 10.1016/j.jbiomech.2008.07.031.
[23] M. M. van der Krogt, S. L. Delp, and M. H. Schwartz, "How robust is human gait to muscle weakness?," *Gait Posture*, vol. 36, no. 1, pp. 113–119, 2012, doi: 10.1016/j.gaitpost.2012.01.017.
[24] T. F. Besier, M. Fredericson, G. E. Gold, G. S. Beaupré, and S. L. Delp, "Knee muscle forces during walking and running in patellofemoral pain patients and pain-free controls," *J. Biomech.*, vol. 42, no. 7, pp. 898–905, 2009, doi: 10.1016/j.jbiomech.2009.01.032.
[25] T. Komura and A. Nagano, "Evaluation of the influence of muscle deactivation on other muscles and joints during gait motion," *J. Biomech.*, vol. 37, no. 4, pp. 425–436, 2004, doi: 10.1016/j.jbiomech.2003.09.022.
[26] A. Forner-Cordero, H. J. F. M. Koopman, and F. C. T. Van Der Helm, "Mechanical model of the recovery from stumbling," *Biol. Cybern.*, vol. 91, no. 4, pp. 212–220, 2004, doi: 10.1007/s00422-004-0508-0.
[27] A. Forner-Cordero, M. Ackermann, and M. De Lima Freitas, "A method to simulate motor control strategies to recover from perturbations: Application to a stumble recovery during gait," in *Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*, 2011, pp. 7829–7832, doi: 10.1109/IEMBS.2011.6091929.
[28] B. I. Prilutsky and R. J. Gregor, "Swing- and support-related muscle actions differentially trigger human walk-run and run-walk transitions.," *J. Exp. Biol.*, vol. 204, no. Pt 13, pp. 2277–87, Jul. 2001, Accessed: Oct. 06, 2019. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/11507111.
[29] A. L. Hof, H. Elzinga, W. Grimmius, and J. P. K. Halbertsma, "Speed dependence of averaged EMG profiles in walking," *Gait Posture*, vol. 16, no. 1, pp. 78–86, 2002, doi: 10.1016/S0966-6362(01)00206-5.
[30] M. G. J. Gazendam and A. L. Hof, "Averaged EMG profiles in jogging and running at different speeds," *Gait Posture*, vol. 25, no. 4, pp. 604–614, 2007, doi: 10.1016/j.gaitpost.2006.06.013.
[31] A. Hreljac, "Determinants of the gait transition speed during human Locomotion : Kinematic factors," *Biomechanics*, vol. 28, no. 6, pp. 669–677, 1995, doi: 0021-9290(94)00120-S [pii].