

# ACUTE EFFECTS OF ANKLE WEIGHT LOADING ON REGIONAL ACTIVITY OF RECTUS FEMORIS MUSCLE AND LOWER-EXTREMITY KINEMATICS DURING WALKING IN OLDER ADULTS

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## Abstract:

Ankle weight loading is a straightforward and easily applicable method to increase the physiological burden during walking in older adults. The current study investigated the acute effects of ankle weight loading on the regional activity of rectus femoris (RF) muscle and lower-extremity joint angles during walking in twenty-nine healthy older adults. Neuromuscular activities of proximal (RFP) and distal (RFD) regions of the RF into averaged rectified values (ARV) and kinematics of left lower extremities were measured during walking with and without ankle weight on a treadmill using surface electromyography (EMG) and motion capture, respectively. Significant differences in normalized ARV between RFP and RFD with ankle weight loading were found in longer periods of the stance phase (30-55 and 5-15%) and swing (70-90 and 95-100%) phases compared to without weight condition (30-40, 50, and 75-85%) ( $p < .05$ ). The hip flexion angle at (10-25 and 60-90%), knee extension at (5%, 15%, 25-35%, and 100%), and ankle dorsiflexion at (30-55% and 75%) of the gait cycle were increased with ankle weight loading more than without it ( $p < .05$ ). Ankle weight loading could change the neuromuscular activity pattern of RFP and improve lower-extremity kinematics during walking in older adults.

**Key words:** *the elderly, ankle weight, rectus femoris, surface electromyography, joint angle*

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## Introduction

The rectus femoris (RF) muscle is activated regionally based on anatomical characteristics during gait phases (Watanabe, Kouzaki, & Moritani, 2014). The most important role of the RF muscle involves its proximal region (RFP) in the swing phase (Watanabe, Kouzaki, & Moritani, 2012). The RF muscle contracts in the stance to swing and terminal swing to extend the knee through concentric and eccentric contractions, respectively (Nardo & Fioretti, 2013). Also, RFP is activated in swing and stance to swing phases to flex the hip joint during gait (Watanabe, et al., 2014). The activity of RFP and the distal regions of RF (RFD) as a bi-articular muscle is affected by various hip and knee joint angles during walking (Savelberg & Meijer, 2004).

Previous studies showed that hip and knee joint range of motion was reduced during gait in older adults (Kirkwood, Gomes, Sampaio, Culham,

& Costigan, 2007). Aging impairs the control of lower-extremity motion during the swing phase and increases the risk of falls (Tanimoto, et al., 2017). Older adults often show reductions in the swing duration and lower-extremity joint motions to prevent falling (Hahn & Chou, 2004). Furthermore, the impairment of lower-extremity kinematics during the swing phase increases the risk of falls in older adults (Kobayashi, Hobara, Matsushita, & Mochimaru, 2014; Watanabe, 2018). The baseline of an age-related decline of lower-extremity motion in healthy older adults is a decrease in knee extensor neuromuscular activity (Reid, et al., 2014).

Neuromuscular activation of the RF muscle to control knee extension and flexion movements within the swing phase is necessary (Tanimoto, et al., 2017). Also, resistance exercise increases electromyography (EMG) activity of RF and vastus lateralis muscles and improves functional activities in young and old subjects (Aagaard, Suetta,

Caserotti, Magnusson, & Kjaer, 2010; Angst, et al., 2015). The researchers reported that resistance exercise increases muscle force production in older adults (Angst, et al., 2015; Carroll, Barry, Riek, & Carson, 2001; Mayer, et al., 2011). Moreover, resistance training recruits muscle fibers and increases motor function in older adults (Bellew, 2002; Carroll, et al., 2001; Mayer, et al., 2011).

Muscle activation between younger and older adults is not the same (Aagaard, et al., 2010). In addition, the age-related declines of lower-extremity muscle activity at different lower-extremity joint angles (Savelberg, et al., 2004) and neuromuscular activities of RFD and RFP in the swing phase are different (Watanabe, et al., 2012). Also, the regional activity of the RF muscle on weight loading while walking in older adults had never been clarified. Therefore, the purpose of the present study was to determine the acute effects of ankle weight loading on the regional RF muscle activity and lower-extremity joint angles during walking in older adults. Regarding previous studies (Watanabe, et al., 2012), the superiority of increased RFP activity in the swing phase could have induced increases in hip flexion and improvements in gait variables in older adults. Thus, we hypothesized that RFP activity would be increased with weight loading and lead to improved older adults' lower-extremity kinematics while walking.

## Methods

### Participants

Twenty-nine healthy older adults (female: 22, male: 7; age:  $67.96 \pm 6.86$  years, body height:  $157.63 \pm 8.48$  cm, body weight:  $53.78 \pm 9.04$  kg, body mass index:  $21.55 \pm 2.38$  kg/m<sup>2</sup>) participated in this study at the Neuromuscular Biomechanics Lab of Chukyo University. The participants gave written informed consent after receiving a detailed explanation about the purposes, potential benefits, and risks of their participation. The older adults who had no medical limitations to the exercise were included as healthy participants and recruited in the local community center in Nagoya city. All the participants were without a history of musculoskeletal or neurological disorders, and none had any exercise limitation advised by a physician. All study procedures were conducted according to the Declaration of Helsinki and the research code of Ethics of Chukyo University, and were approved by the Committee of Human experiment at Chukyo University (2019-002).

### Experimental design

Before starting the experiment, the preferred gait speed of subjects was measured while walking a distance of 10 m normally on flat ground. The subjects walked on a treadmill (TAKEI S-16075,

Takei Kiki Kogyo Co., LTD, Japan) at their preferred gait speed in two trials for 150 s per trial. We used a harness attached to a belt around the participant's waist to prevent falling. Before the start of walking, lower-extremity joint kinematics were captured in a standing position for 5 s. The subjects were familiarized with walking with and without weights on a treadmill for 30 s before starting each trial. Neuromuscular activation of the RF muscle and lower-extremity kinematics were recorded in every trial. The left lower extremity was used to record RF activity and joint kinematics in all trials based on the measurement in the sagittal plane. We recorded muscle activity and joint kinematics 30 s after starting the trial with/ without weight loading. Also, we synchronized EMG and motion capture systems 30 s after recording data. A five-minute rest interval was given between each trial. We applied the weights above the ankle for both legs in the trial with weight loading. The amount of weight loaded was selected from 0.5, 1, 1.5 and 2 kg approximated to 2% of body weight for each participant, determined as the maximum load with which the participant could walk normally on a treadmill. We decreased the weight load and speed of walking when participants experienced difficulty in the familiarization stage of walking on the treadmill.

### Lower-extremity kinematics

A three-dimensional motion capture system with six cameras coordinated by reflective markers on the left lower extremity in the sagittal plane with a sampling rate of 100 HZ was used (Vicon Bonita3, Vicon Motion Systems Ltd., Oxford, United Kingdom). Reflective markers were attached on the left acromion, greater trochanter, lateral femoral condyle, lateral malleolus, fifth metatarsal, toe, and center of the heel. Toe and heel markers were used to detect toe-off and heel contact. Vertical coordinates of the toe and heel were measured before walking in the standing phase to detect heel contact and toe-off timing.

We determined the timing of heel contact as the start of the stance phase minus vertical coordinates and that of toe-off as the start of the swing. The detected coordinates for each marker in the sagittal plane were filtered using a fourth order Butterworth low-pass filter (6 HZ). Also, we synchronized EMG and motion capture data using an infra-radiation light-emitting diode with electrical signals.

### Surface EMG recording

Neuromuscular activity of the RF muscle in the left lower extremity was recorded using surface EMG. The electrodes were attached along a line between ASIS and the midpoint of the superior pole of the patella. The bipolar surface Ag/AgCl (Dual electrodes, 15770 N, Greenway-Hayden Loop,

Scottsdale, AZ 85260, Noraxon, USA, Inc.) electrodes were placed at 20% and 50% from ASIS along the line for RFP and RFD, respectively. The inter-electrode distance was 20 mm and input impedance exceeded 200  $\Omega$ . The sampling rate was 1000 HZ and the signals were filtered with a 4th order Butterworth filter (band width 20-400 Hz) (FA-DL-141, 4 assist, Tokyo, Japan). We shaved the skin on the location of the electrodes, cleaned it with alcohol, and attached the electrodes for RFP and RFD along a line with a common reference electrode on the left iliac crest. Also, we confirmed the electrode locations and muscle activity with isometric contractions of muscles.

### Data analysis

Neuromuscular activity levels of RFP and RFD and hip, knee, and ankle joint values in the sagittal plane were dependent variables. The average rectified values (ARV) of RFP and RFD were analyzed in ten consecutive gait cycles for 120 s after starting each trial for the left lower extremity. ARV of RFP and RFD were normalized by maximum values of ARV during a gait cycle and mean ARV calculated for each 5% of ten gait cycles. The hip joint angle was defined between the greater trochanter of the femur and along the trunk and thigh as hip joint flexion, the knee joint angle was between the lateral condyle and along the thigh and shank as knee flexion, and the ankle joint angle was between the lateral malleolus and condyle of the fifth metatarsal bone as ankle dorsiflexion. The trunk angle was defined between the trunk alignment and vertical axis as the trunk forward flexion angle. Also, the thigh angle was between the anterior side of the thigh and the horizontal axis as the thigh flexion angle.

### Statistical analysis

Data analysis was performed using SPSS (version 20, Tokyo, Japan). We assessed the distribution of data using the Kolmogorov-Smirnov test and analyzed data with nonparametric tests. Also, we compared the acute effect of weight loading on RFP and RFD muscle activity, and hip, knee, ankle, trunk, and thigh joint angles with the Wilcoxon rank sum test between two trials. Every gait cycle was divided into twenty 5% segments and the start and end of every gait cycle were determined as 0 and 100%, respectively. The significance level was set at 0.05.

### Results

There were no significant differences in gait parameters between with and without weight loading, with trials shown in Table 1 ( $p > .05$ ). The average amount of weight loaded was  $0.019 \pm 0.002$  kg and the preferred gait speed was  $4.68 \pm 0.62$  (km/h). The mean ARV of RFP in 70% of the gait cycle with was greater than without loading ( $p < .05$ ) (Fig. 1). There were significant differences between trials in mean ARV of RFP at 10, 20, and 95% of the gait cycle ( $p < .05$ ) (Fig. 1) and mean ARV of RFD at 20-30 and 75% ( $p < 0.05$ ) (Fig. 2). Mean ARV of RFP with weight loading was more than that of RFD at

Table 1. Gait parameters during walking with and without weight loading

|                           | With loading      | Without loading    | p-value |
|---------------------------|-------------------|--------------------|---------|
| Cadence (Step number/min) | 127.33 $\pm$ 7.47 | 125.52 $\pm$ 18.68 | 0.059   |
| Toe off timing (%)        | 63.30 $\pm$ 1.62  | 63.70 $\pm$ 2.02   | 0.125   |

Note.  $p < .05$  is significant.

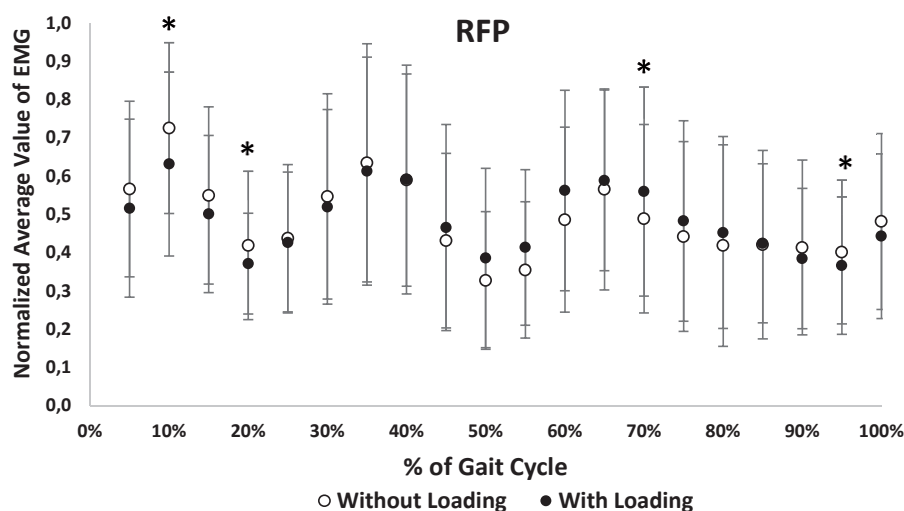


Fig. 1. Mean normalized absolute rectified values (mean and standard deviation) of the proximal region of the rectus femoris (RFP) muscle during walking with and without weight loading. \* indicates  $p < .05$ .

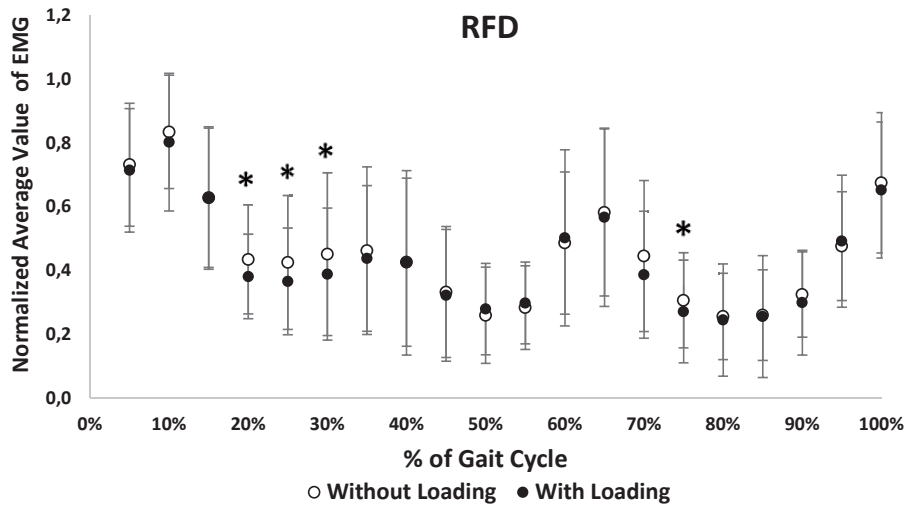


Fig. 2. Mean normalized absolute rectified values (mean and standard deviation) of the distal region of the rectus femoris (RFD) muscle during walking with and without weight loading. \* indicates  $p < .05$ .

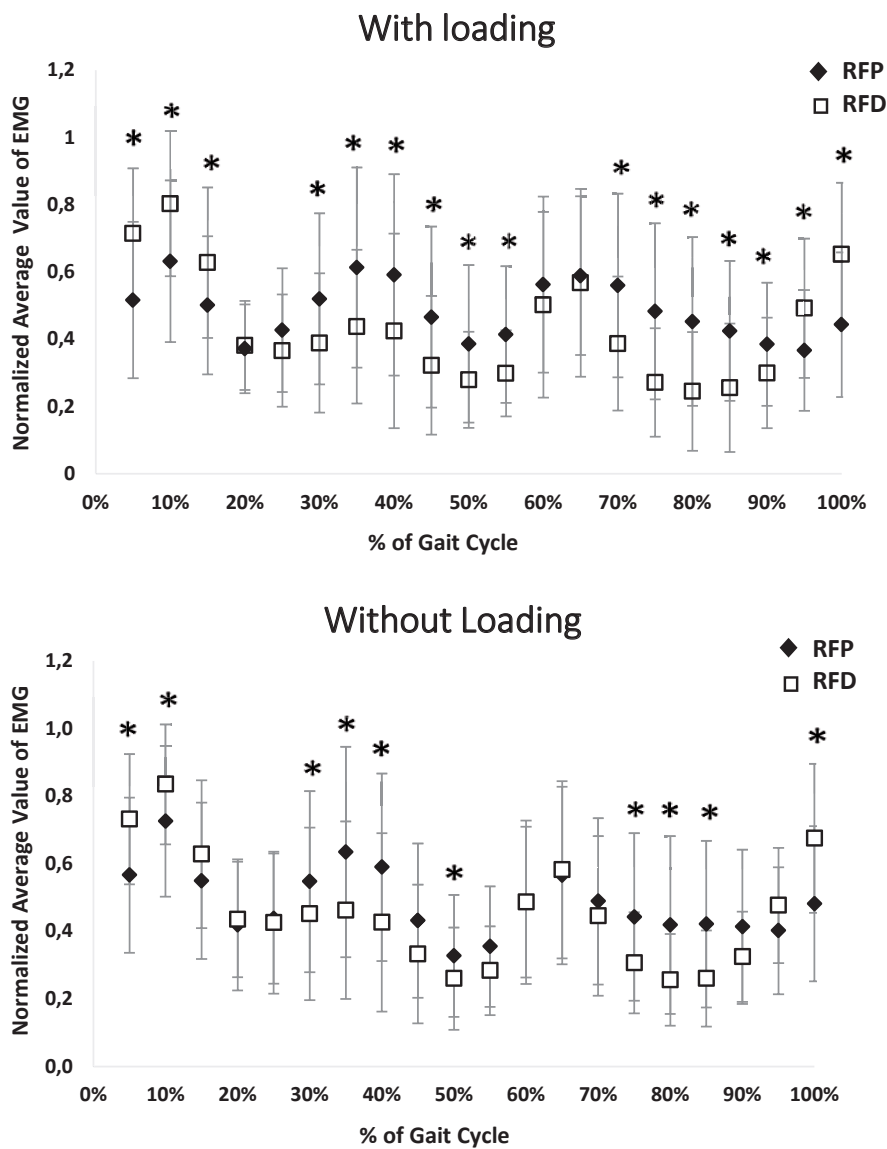


Fig. 3. Mean normalized absolute rectified values (mean and standard deviation) of the proximal (RFP) and distal (RFD) regions of the rectus femoris muscle during walking with and without weight loading. \* indicates  $p < .05$ .

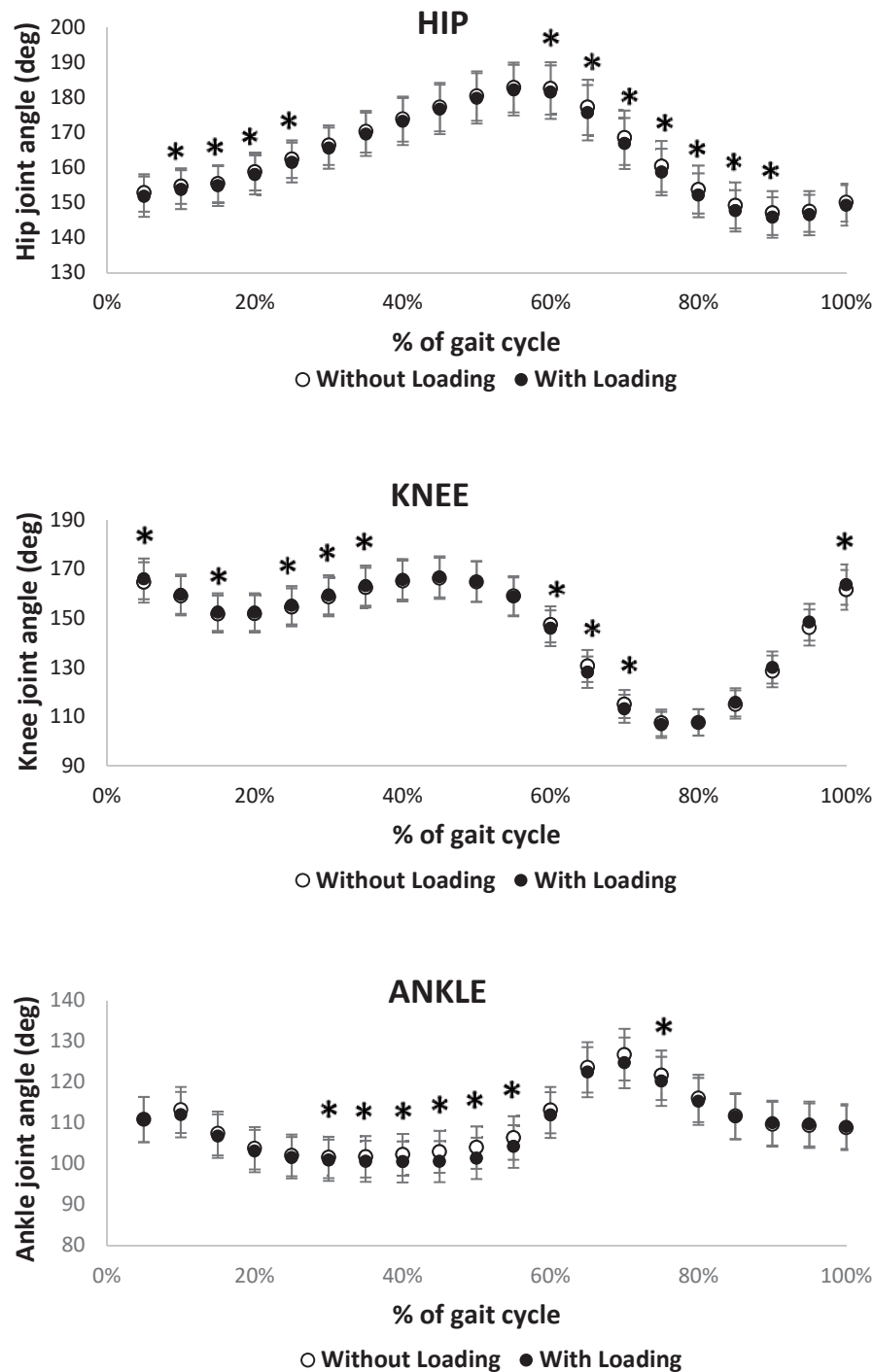


Fig. 4. Hip, knee, and ankle joint range of motion (degrees) in the sagittal plane during walking with and without loading. \* indicates  $p < .05$ .

30-55 and 70-90% of the gait cycle ( $p < .05$ ) (Fig. 3). Mean ARV of RFP at 30-40, 50 and 75-85% of the gait cycle was more than that of RFD during walking without weight loading ( $p < .05$ ) (Fig. 3). Mean ARV of RFD with weight loading was more than that of RFP at 5-15 and 95-100% ( $p < .05$ ) (Fig. 3). Mean ARV of RFD without weight loading was more than that of RFP at 5-10 and 100% ( $p < .05$ ) (Fig. 3).

There were significant differences in knee joint angles at 25-35 and 60-70, 5, 15 and 100% of the gait cycle between the trials ( $p < .05$ ) (Fig. 4). There were significant differences in hip joint angles at 10-25 and 60-90% and ankle joint angles at 30-55 and 75% between the trials ( $p < .05$ ) (Fig. 4). Hip joint angle ( $p = .32$ ) and thigh angle ( $p = .13$ ) changes were not significant between the trials. Also, trunk angle changes were significant between the trials ( $p = .013$ ).

## Discussion and conclusions

The increased resistance enhances the RF muscle activity level in the swing phase during walking on a treadmill in young subjects (Klarner, Blouin, Carpenter, & Lam, 2013). According to our results, the acute use of weight loading during walking could not increase the neuromuscular activity of RFP except at 1% of the gait cycle (Fig. 1). We suggest that the increased activity of RFP at 1% of the gait cycle is negligible and it is inappropriate to come to any conclusion regarding the acute effect of weight loading on RFP activity. The weight load approximating to 1% of the body weight increased gluteus medius activity and improved the quality of walking (Hwang, et al., 2017). Gluteus medius muscle activity with weight loading of 2% was lower than that with 1%, although trunk stabilizer muscle activity was increased with a loading of 2% (Lee, 2013). Thus, there was no positive relationship between increased resistance and muscle activity; it was dependent on the muscle function and direction of the applied resistance (Lee, 2013; Simpson, Munro, & Steele, 2011). In our study, the load was set at 2% of the body weight based on the capability of older adults to walk normally on a treadmill. We suggest that the load was too low to induce significant acute effects on RF muscle activity and, applying such a large load acutely during walking on a treadmill would be difficult for older adults.

Previous findings showed that RFP activity is greater than RFD activity during hip flexion, and RFD activity is greater than RFP activity during knee extension (Watanabe, et al., 2012). Our findings confirmed these results with more detailed data, whereby RFP activities from mid-stance until toe-off (30-55%) and early swing to terminal swing (70-95%) with weight loading were greater than those of RFD (Fig. 3). Neuromuscular activity of RFP without weight loading in pre-swing and mid-swing (30-40, 55, 75-85%) was greater than that of RFD (Fig. 3). It was shown that weight loading more markedly increases RFP activity related to RFD on contraction during gait phases. In pre-swing, when the elevation of the lower extremity demanded marked hip and knee joint flexion, there was no significant difference between the RFP and RFD activity levels (Whittle, 2008). The central nervous system primarily activates the quadriceps femoris muscle during knee flexion in pre-swing and, along with it, quadriceps muscle activity is reduced to extend the knee (Shields, et al., 2005).

Moreover, RFD activity levels with weight loading were greater than those of RFP at the start and end of the gait cycle (0-15 and 95-100%) (Fig. 3). The main function of the RF muscle is the transmission of the lower extremity from stance to swing via changing RFP and RFD contractions. Therefore, it is not possible to completely separate the function

of the two regions of the RF muscle. Regarding the importance of the effect of the RF muscle function on the two joints, RFP and RFD activity levels for both joints based on the demand during gait phases should be optimized. RFP and RFD activity levels at more specific percentages of gait phases were increased (Fig. 3), and weight loading could coordinate RF contraction regionally based on the demand in gait phases. Miyamoto, Wakahara, and Kawakami (2012) reported that RFP and RFD activities were increased similarly during knee extension, but RFD was not completely activated during hip flexion (Miyamoto, et al., 2012). Our findings support these results and reveal that RFP was higher than RFD neuromuscular activity in larger percentages of the gait cycle from early swing to terminal swing and mid-stance until toe-off. We clarified that RFP and RFD activity levels at the beginning and end of the gait cycle during knee extension were not similar in either phase. Therefore, our results did not support the presence of similar increases in RFP and RFD activities during knee extension.

From a biomechanical perspective, hip joint flexion should be increased in mid-swing to elevate the lower extremity and move the swinging limb forward in relation to the stance limb accompanied by increased knee joint flexion (Whittle, 2008). In our study, the hip flexion angle with loading was increased in the stance (10-25%) and swing (60-90%) phases (Fig. 4). The maximum performance demand on the hip and knee extensor muscles during walking was increased with aging, and older adults showed higher rates of falling and injury (Samuel, Rowe, & Nicol, 2013). Our results showed that weight loading increased the hip flexion angle in specific percentages of gait phases to guide the body forward and elevate the lower extremity. After further analysis of thigh and trunk joint angle data, we found that the thigh angle was not increased but that the trunk flexion angle was increased with weight loading. Therefore, we suggest that older adults adjusted their specific kinematics with activating hip flexors such as the psoas and abdominal muscles. Although we did not measure the activity of these muscles, the increased trunk flexion angle suggests the contribution of hip flexors to the increased hip flexion angle. Thus, the increased hip flexion angle that we observed could have been the result of the contributions of the abdominal and psoas muscles.

Changing the knee and hip joint angles during lower-extremity movements affects the activation of synergist muscles of the trunk and pelvis (Narouei, Imai, Akuzawa, Hasebe, & Kaneoka, 2018). When hip and knee muscles are weak, synergist muscles will be activated during walking to control lower-extremity movement and compensate for function deficiency (Whittle, 2008). RFP was activated in older adults to a lesser extent than in young subjects

during walking (Watanabe, Kouzaki, & Moritani, 2016). Although we instructed participants to walk normally and used proper feedback to control the normal kinematics, it was not possible to completely avoid muscle co-activation. The participants flexed their trunk to prevent instability and falling during walking with weight loading. Therefore, we suggest that future studies should evaluate abdominal muscle activity during walking after an intervention period with weight loading. The knee extension angle was increased with weight loading in the stance phase (25-35%), and at the beginning and end of the gait cycle (5, 15 and 100%) (Fig. 4). Resistance training improved the dynamic control and stability of the knee joint during gait (Shields, et al., 2005). Our results confirmed that weight loading improves knee extension during walking in older adults. These results are useful to prevent falling and reduce limitations of knee movement for application in rehabilitation clinics. Also, the ankle dorsiflexion angle with weight loading was increased during the stance and swing phases (30-55 and 75%) (Fig. 4). Although we applied weight loading above the ankle, ankle dorsiflexion was increased to improve foot clearance in toe-off. These results showed that weight loading improves lower-extremity joint kinematics during functional movement.

There were some potential limitations in this study. Greater levels of weight loading may provide clearer results on RF activity (Washabaugh, Augenstein, & Krishnan, 2020; Simpson, et al., 2011), but there were limitations on the weight that could be loaded for older adults during walking on the treadmill. Also, the left lower extremity was

assessed in the sagittal plane in twenty-nine older adults; no determination of the differences between the right and left lower extremities and the small sample size could be the other limitations in our study. Moreover, our purpose was to determine the regional activity of RFP and RFD. We did not evaluate the neuromuscular activity of the psoas and trunk muscles, despite the fact these muscles flex the hip during walking. Psoas muscles are the main hip flexors in the swing phase and abdominal muscles are the trunk stabilizers during walking. Due to the contribution of different muscle groups to walking, focusing on changes in the activity of a single muscle activity is difficult. We recommend that future studies investigate the regional activity of the RF muscle regarding co-activation between muscles during walking.

Ankle weight loading increased the neuromuscular activity level of RFP relative to that of RFD in stance and swing phases and RFD activity level relative to that of RFP in early stance and terminal swing phases in more specific percentages of gait phases. Also, ankle weight loading increased knee extension and ankle dorsiflexion angles. Neuromuscular activation and biomechanical changes in our study revealed that weight loading could be useful to guide rehabilitation treatment methods that involve applying ankle weight loading during walking prescribed to improve RF regional activity and range of motion of the knee. We recommend more investigations on the effect of using ankle weight loading in functional rehabilitation programmes since it may prevent limitation of motion and reduce walking impairments in older adults.

## References

- Aagaard, P., Suetta, C., Caserotti, P., Magnusson, S.P., & Kjaer, M. (2010). Role of the nervous system in sarcopenia and muscle atrophy with aging: Strength training as a countermeasure. *Scandinavian Journal of Medicine and Science in Sports*, 20, 49-64. doi: 10.1111/j.1600-0838.2009.01084
- Angst, F., Kaufmann, M., Benz, Th., Nehrer, S., Aeschlimann A., & Lehmann, S. (2015). Quadriceps performance under activation of foot dorsal extension in healthy volunteers: An interventional cohort study. *BMC Musculoskeletal Disorders*, 16, 340. doi: 10.1186/s12891-015.0774-0.
- Bellew, J.W. (2002). The effect of strength training on control of force in older men and women. *Aging Clinical and Experimental Research*, 14, 35-41. doi: 10.1007/BF03324415
- Carroll, T.J., Barry, B., Riek, S., & Carson, R.G. (2001). Resistance training enhances the stability of sensorimotor coordination. *Proceedings of the Royal Society-Biological Sciences*, 268, 221-227. doi: 10.1098/rspb.2000.1356
- Hahn, M.E., & Chou, L.S. (2004). Age-related reduction in sagittal plane center of mass motion during obstacle crossing. *Journal of Biomechanics*, 37, 837-44. doi: 10.1016/j.jbiomech.2003.11.010
- Hwang, J.W., Lee, S.K., Park, J.S., Ahn, S.H., Lee, K.J., & Lee, S.J. (2017). The effects of ankle weight loading on the walking factors of adults without symptoms. *Journal of Exercise Rehabilitation*, 13, 425-429. doi: 10.12965/jer.1734954.477
- Kirkwood, R.N., Gomes, H.D.A., Sampaio, R.F., Culham, E., & Costigan, P. (2007). Biomechanical analysis of hip and knee joints during gait in elderly subjects. *Acta Ortopédica Brasileira*, 15, 267-271
- Klarner, T., Blouin, J.S., Carpenter, M.G., & Lam, T. (2013). Contributions to enhanced activity in rectus femoris in response to Lokomat-applied resistance. *Experimental Brain Research*, 225, 1-10. doi: 10.1007/s00221-012-3345-8

- Kobayashi, Y., Hobara, H., Matsushita, S., & Mochimaru, M. (2014). Key joint kinematic characteristics of the gait of fallers identified by principal component analysis. *Journal of Biomechanics*, *47*, 2424-2429. doi: 10.1016/j.jbiomech.2014.04.011
- Lee, S.K. (2013). The effect of a vertical load on gluteus medius activity and gait characteristics during walking. *Journal of Physical Therapy Science*, *25*, 1397-1399. doi: 10.1589/jpts.25.1397
- Mayer, F., Scharhag-Rosenberger, F., Carlsohn, A., Cassel, M., Muller, S., & Scharhag, J. (2011). The intensity and effects of strength training in the elderly. *Deutsches Ärzteblatt International*, *108*, 359-364. doi: 10.3238/arztebl.2011.0359
- Miyamoto, N., Wakahara, T., & Kawakami, Y. (2012). Task dependent inhomogeneous muscle activities within the bi articular human rectus femoris muscle. *Plos One*, *7*, 34269. doi: 10.1371/journal.pone.0034269
- Nardo, F.D., & Fioretti, S. (2013). Statistical analysis of surface electromyographic signal for the assessment of rectus femoris modalities of activation during gait. *Journal of Electromyography and Kinesiology*, *23*, 56-61. doi: 10.1016/j.jelekin.2012.06.011
- Narouei, S., Imai, A., Akuzawa, H., Hasebe, K., & Kaneoka, K. (2018). Hip and trunk muscles activity during Nordic hamstring exercise. *Journal of Exercise Rehabilitation*, *14*, 231-238. doi: 10.12965//jer.1835200.600
- Reid, K.F., Pasha, E., Doros, G., Clark, D.J., Patten, C., Phillips, E.M., et al. (2014). Longitudinal decline of lower extremity muscle power in healthy and mobility-limited older adults: Influence of muscle mass, strength, composition, neuromuscular activation and single fiber contractile properties. *European Journal of Applied Physiology*, *114*, 29-39. doi: 10.1007/s00421-013-2728-2
- Savelberg, H.H.C.M., & Meijer, K. (2004). The effect of age and joint angle on the proportionality of extensor and flexor strength at the knee joint. *Journal of Gerontology*, *59*, 1120-1128. doi: 10.1093/gerona/59.11.1120
- Shields, R.K., Madhavan, S., Gregg, E., Leitch, J., Petersen, B., Salata, S., et al. (2005). Neuromuscular control of the knee during a resisted single-limb squat exercise. *The American Journal of Sports Medicine*, *33*, 1520-1526. doi: 10.1177/0363546504274150
- Simpson, K.M., Munro, B.J., & Steele, J.R. (2011). Backpack load affects lower limb muscle activity patterns of female hikers during prolonged load carriage. *Journal of Electromyography and Kinesiology*, *21*, 782-788. doi: 10.1016/j.jelekin.2011.05.012
- Samuel, D., Rowe, P., & Nicol, A. (2013). The functional demand (FD) placed on the knee and hip of older adults during everyday activities. *Archives of Gerontology and Geriatrics*, *57*, 192-197. doi: 10.1016/j.archger.2013.03.003
- Tanimoto, K., Takahashi, M., Tokuda, K., Sawada, T., Anan, M., & Shinkoda, K. (2017). Lower limb kinematics during the swing phase in patients with knee osteoarthritis measured using an inertial sensor. *Gait Posture*, *57*, 236-240. doi: 10.1016/j.gaitpost.2017.06.017
- Washabaugh, E.P., Augensteina, T.E., & Krishnan, C. (2020). Functional resistance training during walking: Mode of application differentially affects gait biomechanics and muscle activation patterns. *Gait Posture*, *75*, 129-136. doi: 10.1016/j.gaitpost.2019.10.024
- Watanabe, K., Kouzaki, M., & Moritani, T. (2012). Task-dependent spatial distribution of neural activation pattern in human rectus femoris muscle. *Journal of Electromyography and Kinesiology*, *22*, 251-258. doi: 10.1016/j.jelekin.2011.11.004
- Watanabe, K., Kouzaki, M., & Moritani, T. (2014). Regional neuromuscular regulation within human rectus femoris muscle during gait. *Journal of Biomechanics*, *47*, 3502-3508. doi: 10.1016/j.jbiomech.2014.09.001
- Watanabe, K., Kouzaki, M., & Moritani, T. (2016). Regional neuromuscular regulation within human rectus femoris muscle during gait in young and elderly men. *Journal of Biomechanics*, *49*, 19-25. doi: 10.1016/j.jbiomech.2015.11.010
- Watanabe, K. (2018). Relationship between toe clearance strategy and regional regulation of rectus femoris muscle during swing phase in prolonged walking in young and older adults. *Frontiers in Physiology*, *9*, 1274. doi: 10.3389/fphys.2018.01274
- Whittle, M.W. (2008). *Gait analysis: An introduction*. Britain: Butterworth Heinemann.

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