Asian J of Human Services Vol.25 217-224, 2023 DOI: 10.14391/ajhs.25.217

Asian Journal of Human Services

Journal homepage: https://www.ashs-human.net/international-journal/ajhs/
Online ISSN: 2188-059X / Print ISSN: 2186-3350

SHORT PAPER

Influence of Kyphosis Posture on Lower Extremity Joint Moment Power during Walking

Chihiro NIIBO[1)*], Ryouta NAGAHAMA[1)], Hidefumi FUKUDA[1)], Hiroshi KATOH[2)]

1) Kohshinkai Ogura Hospital, Japan
2) Yamagata Prefectural University of Health Sciences, Japan

ABSTRACT

This study was conducted in order to investigate the effect of kyphosis alignment during walking on lower extremity joint power at the stance phase in healthy participants. The composite components of lower extremity joint power in three axial directions were calculated by using a three-dimensional motion analyzer in 15 healthy adult male participants. Two postural conditions, viz., a normal posture (normal posture) and a kyphosis posture with a brace (kyphosis posture), were used. The average and maximum values of joint power in the early, mid, and terminal phases of stance were compared. In the early stance phase, the average, and maximum values of lower extremity joint power were significantly different at the hip and knee joints. The hip joint power was significantly different in the terminal stance phase. The values of hip and knee joint power were significantly different in the terminal stance phase. The workload and work rate increased with hip power in the normal posture and with knee power in the kyphosis posture.

Keywords: Joint power, kyphosis posture, walking

* Chihiro NIIBO: chihiro.niibo.pt@gmail.com

Received: 2023/04/05 ; *Revised*: 2023/07/27 ; *Accepted*: 2023/08/27 ; *Published*: 2023/10/30

1. Introduction

The joint moment power (joint power) in walking motion is defined as the product of joint moment and joint angular velocity and is a mechanical parameter which indicates instantaneous force, i.e., the work rate. Furthermore, the joint power can infer the contraction pattern of the muscle group which exerts the joint movement. Positive work of joint power is the work done during a concentric contraction, whereas negative work is the work done during an eccentric contraction. Hence, joint power is considered a clinically important index which can estimate the positive work (energy generation) and negative work (energy absorption) of each joint during walking[1)].

There are several previous studies on healthy participants and patients with lower extremity joint diseases. The effects of walking speed[2)] and differences in conditions such as level ground and slopes[3)] were considered in the case of healthy participants. Studies were focused on during walking in patients with knee osteoarthritis and patients after total hip replacement surgery as regards patients with lower extremity joint diseases[4,5)]. It has been reported to affect distal adjacent joints in addition to decreased joint power due to functional impairment of the affected articulation[5,6)]. In contrast, it has been reported that malalignment of the proximal trunk and postural changes in the standing posture exert a significant impact on lower extremity alignment[7,8,9)]. A study focusing on differences in standing postures reported that when the trunk is tilted forward (trunk flexion) in healthy participants, the hip joint flexion angle increases as a compensatory movement[10)], the hip extension moment increases[11)], and the activity time is prolonged[12)] in the early stance phase during walking. The peak moment and angular impulse of hip flexion have been reported to decrease during the mid-stance phase[13)]. In the knee joint, the trunk flexion affects the knee joint moment during the entire stance phase, which increases the activity of the lateral hamstrings[14)]. These studies have provided some insights into the effects of postural changes on lower extremity joints. However, there are no studies which have analyzed in detail the characteristics of joint power dynamics and the contraction patterns of muscle groups.

Therefore, this study was conducted to investigate the effect of kyphosis alignment during walking on lower extremity joint power at the stance phase in healthy participants.

2. Methods

2.1. Participants

This study included 15 healthy men (mean ± standard deviation; age 26.40 ± 6.49 years; height 169.20 ± 4.49 cm; weight 67.40 ± 10.14 kg) with no previous orthopedic, neurological, respiratory, or circulatory diseases, excluding those with a kyphosis alignment in a quiet standing posture. The walking ability of the participants was 1.58 ± 0.17 m/sec at 10 m walking speed and 283.98 ± 29.71 m at 3 min walking distance. Participants had no history of falls or symptoms of back or knee pain.

This study was approved by the Ethics Committee of the Kohshinkai Ogura Hospital (2018-A1). The participants were informed about the outline of the study in the research consent form, and their consent to participate in the study and signatures on the consent form were obtained. Participants were invited to participate in the study by displaying posters at the campuses of the research institutions.

2.2. Motion task

The task was walking, and the following two postural conditions were used: a posture with no specific regulations (normal posture) and a kyphosis posture with a brace (kyphosis posture).

2.3. Measurement

A trunk orthosis, a Vicon MX-T three-dimensional motion analysis system with ten infrared cameras (Vicon Motion Systems Ltd., Oxford, UK), and six force plates (AMTI Inc., Watertown, MA, USA) were used for measurements. First, a three-point fixation of the sternum, abdomen, and back was performed by using the orthosis in maintaining a thoracolumbar kyphosis position. The brace angle was set to 160° (Fig 1), and the brace was implemented after confirming that the angle between the lines connecting C7-Th9 and Th9-S3 showed a kyphosis posture of 25°–30°. The angles of the neck, hip, and knee joints were not set but were left to the subject's discretion to facilitate maintaining the kyphosis posture. Second, 33 reflective markers were placed on body markers. The sampling rate was 100 Hz, and the marker coordinate data and the force plates were processed by applying a low-pass filter at 6 and 10 Hz, respectively. An eight-link segmental model was created by using Body Builder (Vicon) from the obtained marker coordinate data and the ground reaction force data[15,16)], and the composite components of the lower extremity joint power in the three axial directions were calculated. The value of joint power was normalized with respect to participant body mass.

The participants walked on a 10-m walkway which was set up with an acceleration interval at the start of walking and a deceleration interval of 2 m at the end, and measurements were taken after three to five practice sessions for each condition. The walking speed with a stride length of 0.7 m and a walking rate of 117 steps/min was specified. The walking rate was set to the sound of an electronic metronome ME-110 (YAMAHA Inc., Shizuoka, Japan), and markers were placed on the side of the walkway to serve as a guide.

2.4. Data analyses

The parameters hip, knee, and ankle joint power during the stance phase of the left lower extremity were analyzed. The stance phase was identified as the point at which the left plantar foot touched the floor and the vertical ground reaction force was ≥20 N[14,17)], and the stance end point was identified when the vertical ground reaction force was <20 N. Then, the left lower extremity stance period time was time-normalized and classified into four periods, viz., 0%–20% (early stance period), 21%–52% (mid-stance period), 53%–83% (terminal stance period), and 84%–100%, and the mean, and maximum values of joint power were calculated for each period.

2.5. Statistical analyses

The average and maximum (absolute) values of joint power in the early, mid, and terminal phases of stance were compared between normal and kyphosis postures. Statistical analysis was performed by using R4.0.2. The Shapiro–Wilk test was used in examining normality, and the two-sample t-test, and Wilcoxon's signed-rank test were used to examine differences between groups. P values of <0.05 were considered to be statistically significant.

3. Results

The average lower extremity joint power was significantly different at the hip and knee joints in the early stance phase between the normal posture and kyphosis posture. The ankle joint power was significantly different in the mid-stance phase. The hip joint power was significantly different in the terminal stance phase. The maximum lower extremity joint power was significantly different at the hip and knee joints in the early stance phase between the normal posture and kyphosis posture. The knee joint power was significantly different in the mid-stance phase. The values of hip and knee joint power were significantly different in the terminal stance phase (Table 1).

4. Discussion

Approximately 60% of the gait cycle is the stance phase[18)], and approximately 40% is the swing phase. Moreover, approximately 10% of the beginning and end of the stance phase is double-leg support, and the remaining is single-leg support. Winter[19)] reported that lower extremity extensor activity is necessary for push-off against lower extremity flexion during the stance phase.

The hip joint power was high in the normal posture, and the knee joint power was high in the kyphosis posture in the early stance phase based on the results of this study. The phase from the initial contact to the loading response is a period of shock-absorbing action, which is primarily controlled by the gluteus maximus and rectus femoris muscles, the foot rocker function, and knee joint flexion motion. Particularly, the function of the eccentric contraction of the hip extensor muscle is important, and it is believed that this function is difficult to exert in the kyphosis posture although the gluteus maximus muscle generates greater support[20)] and braking force in the early stages of stance. Furthermore, even at the maximum value in the early stance phase, the knee joint power was higher than that in the normal posture. The body's center of gravity is shifted backward to the base of support[21)], which requires more activity of the knee extensor muscles in the standing posture with kyphosis. Similarly, it has been reported that the knee joint extension moment increases in the early stance phase during walking[11)]. This suggests that the kyphosis posture increases the workload and work rate of the knee joint extensor muscles throughout the early stance phase.

The ankle joint power was high in the kyphosis posture in the mid-stance phase. The body's center of gravity moves forward while receiving the load from the load response period, and the muscle activity of the ankle increases in the mid-stance period. Stephen et al. reported that the ankle plantar flexion moment increases as the trunk flexion increases. Moreover, the angle change ratio of ankle dorsiflexion increases in the trunk flexion position[10)], which implies that the dorsiflexion motion that occurs after the load is transferred, is more likely to occur in the kyphosis posture. Kluger et al.[11)] reported that the ankle joint power increases during this phase, which suggests that in the kyphosis posture, the ankle joint power influences the control of dorsiflexion that occurs after the load is transferred.

The hip joint power was high in the normal posture, and the maximum knee joint power was high in the kyphosis posture in the terminal stance phase. In the terminal stance phase, eccentric contraction by the hip flexors is vital to gently shift the body's center of gravity, which reaches its apex in the mid-stance phase anteriorly and downward. Moreover, the stretching of the hip flexors (eccentric contraction) in the terminal stance phase helps in storing energy and providing support for subsequent preswing and initial swings[22)]. A study on patients who underwent total hip arthroplasty reported that a decrease in hip joint power in the terminal stance phase is probably accompanied by an increase in ankle joint power, which may be compensated for by the action of the ankle plantar flexor muscles[5)]. No compensation by the ankle joint power was observed, but it was considered that eccentric contraction by the hip joint was increased in the normal posture, and the braking action was more likely to be exerted in this study.

This study is unique as it shows that postural changes affect lower extremity joint power during gait. Hip flexion contracture has been described as a result of decrease in the peak hip extension moment[23)], and kyphosis posture may be susceptible to this effect. Passive joint moments during the mid to terminal stance phase have been reported to occur in the hip flexors, which contributes to a subsequent reduction in the active energy of the individual[24)]. It has been suggested that exercise and movement guidance promotes normal posture and are enhances the hip power.

A limitation of this study is that the physiological kyphosis posture is not exhibited by elderly individuals; hence, it may differ from the compensatory and compensatory relationships of the ankle and hip joints described in previous studies. Moreover, it is unclear whether the thoracic spine or the lumbar spine is the cause since the entire spine is in a kyphosis posture. It is important to investigate the influences of age and participants who actually have kyphosis in the future.

5. Conclusion

This study investigated the effect of kyphosis posture on lower extremity joint power during walking in healthy participants. The hip joint power was high in the normal posture and the knee joint power was high in the kyphosis posture in the early and terminal phases of stance. The workload and work rate of the knee joint increased in the kyphosis posture.

Table 1. Joint moment power

Phase	Early-stance		Mid-stance		Terminal-stance	
Joint power (W/kg)	Normal	Kyphosis	Normal	Kyphosis	Normal	Kyphosis
Hip ave	0.66±0.22	0.46±0.18*	0.4±0.15	0.36±0.38	0.63±0.17	0.44±0.16**
Hip max	1.26±0.46	0.89±0.32*	0.87±0.32	0.63±0.49	0.91±0.21	0.79±0.26*
Knee ave	0.55±0.18	0.87±0.3**	0.29±0.17	0.34±0.2	0.18±0.07	0.24±0.11
Knee max	0.99±0.37	1.55±0.63**	0.57±0.35	0.9±0.61**	0.39±0.21	0.71±0.31**
Ankle ave	0.36±0.16	0.36±0.13	0.2±0.07	0.35±0.21**	0.6±0.19	0.64±0.28
Ankle max	0.69±0.3	0.71±0.24	0.42±0.17	0.56±0.22	2.01±0.75	2.16±1.25

Note: Data are expressed as mean ± SD. *p<0.05, **p<0.01.

Hip ave, average of hip joint power; Hip max, maximum of hip joint power; Knee ave, average of Knee joint power; Knee max, maximum of Knee joint power; Ankle ave, average of Ankle joint power; Ankle max, maximum of Ankle joint power.

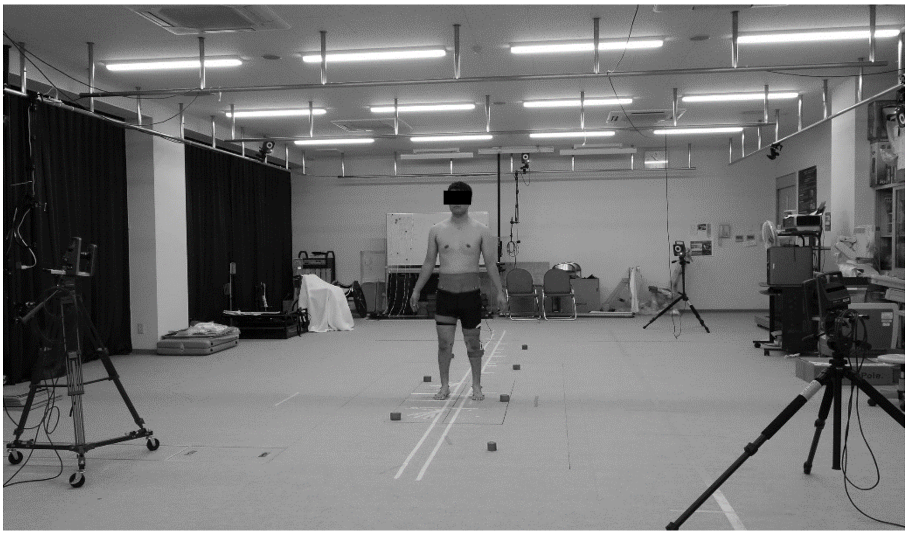

Fig 1. Measurement Environment

Fig 2. Set the angle between A-B and B-C to 160°

Adjust the length of A-B and B-C according to the target person.

Acknowledgements

The authors thank all individuals who participated in this study. The authors would like to thank Enago (https://www.enago.jp/) for the English language review.

References

1) Winter DA. *Biomechanics and Motor Control of Human Movement. Chap 6. In: Mechanical Work, Energy, and Power*, 4th edition. 2009, John Wiley & Sons. Hoboken.
2) Pimentel RE, Feldman JN, Lewek MD & Franz JR. Quantifying mechanical and metabolic interdependence between speed and propulsive force during walking. *Front Sports Act Living*. 2022, 4, 1-12. DOI: 10.3389/fspor.2022.942498.
3) DeVita P, Helseth J & Hortobagyi T. Muscles do more positive than negative work in human locomotion. *J Exp Biol*. 2007, 210(Pt 19), 3361-3373. DOI: 10.1242/jeb.003970.
4) Resende RA, Fonseca ST, Silva PL, Magalhães CM & Kirkwood RN. Power at hip, knee and ankle joints are compromised in women with mild and moderate knee osteoarthritis. *Clin Biomech*. 2012, 27(10), 1038-1044. DOI: 10.1016/j.clinbiomech.2012.08.001.
5) Tateuchi H, Tsukagoshi R, Fukumoto Y, Oda S & Ichihashi N. Dynamic hip joint stiffness in individuals with total hip arthroplasty: relationships between hip impairments and dynamics of the other joints. *Clin Biomech*. 2011, 26(6), 598-604. DOI: 10.1016/j.clinbiomech.2011.01.014.
6) Lewis CL & Ferris DP. Walking with increased ankle pushoff decreases hip muscle moments. *J Biomech*. 2008, 41(10), 2082-2089. DOI: 10.1016/j.jbiomech.2008.05.013.
7) Severijns P, Moke L, Overbergh T, Beaucage-Gauvreau E, Ackermans T, Desloovere K, et al. Dynamic sagittal alignment and compensation strategies in adult spinal deformity during walking. *Spine J*. 2021, 21(7), 1059-1071. DOI: 10.1016/j.spinee.2021.02.017.
8) Otayek J, Bizdikian AJ, Yared F, Saad E, Bakouny Z, Massaad A, et al. Influence of spino-pelvic and postural alignment parameters on gait kinematics. *Gait Posture*. 2020, 76, 318-326. DOI: 10.1016/j.gaitpost.2019.12.029.
9) Sakuma T & Ae M. Biomechanics effects of the trunk inclination on gait. *Biomechanism Gakkaishi* (in Japanese). 2010, 34(4), 325-332.
10) Saha D, Gard S & Fatone S. The effect of trunk flexion on able-bodied gait. *Gait Posture*. 2008, 27(4), 653-660. DOI: 10.1016/j.gaitpost.2007.08.009.
11) Kluger D, Major MJ, Fatone S & Gard SA. The effect of trunk flexion on lower-limb kinetics of able-bodied gait. *Hum Mov Sci*. 2014, 33, 395-403. DOI: 10.1016/j.humov.2013.12.006.
12) Leteneur S, Gillet C, Sadeghi H, Allard P & Barbier F. Effect of trunk inclination on lower limb joint and lumbar moments in able men during the stance phase of gait. *Clin Biomech*. 2009, 24(2), 190-195. DOI: 10.1016/j.clinbiomech.2008.10.005.
13) Lewis CL & Sahrmann SA. Effect of posture on hip angles and moments during gait. *Man Ther*. 2015, 20(1), 176-182. DOI: 10.1016/j.math.2014.08.007.
14) Alghamdi W & Preece SJ. How does normal variability in trunk flexion affect lower limb muscle activity during walking?. *Hum Mov Sci*. 2020, 72, 1-10. DOI: 10.1016/j.humov.2020.102630.

15) Kito N, Shinkoda K, Yamasaki T, Kanemura N, Anan M, Okanishi N, et al. Contribution of knee adduction moment impulse to pain and disability in Japanese women with medial knee osteoarthritis. *Clin Biomech*. 2010, 25(9), 914-919. DOI: 10.1016/j.clinbiomech.2010.06.008.

16) Kito N, Shinkoda K, Kanemura N, Anan M, Yamazaki T, Ishi S, et al. External Knee Adduction Moment during Stepping in Subjects with Medial Knee Osteoarthritis is Correlated with Pain and Disability. *Rigakuryoho Kagaku* (in Japanese). 2008, 23(5), 633-640.

17) Preece SJ & Alghamdi W. The effect of increasing trunk flexion during normal walking. *Gait Posture*. 2021, 83, 250-255. DOI: 10.1016/j.gaitpost.2020.10.021.

18) Hase K & Munkhdelger D. Basic Viewpoint and Procedure of Gait Analysis. *Jpn J Rehabil Med* (in Japanese). 2021, 58(2), 114-120.

19) Winter DA. Overall principle of lower limb support during stance phase of gait. *J Biomech*. 1980, 13(1), 923-927. DOI: 10.1016/0021-9290(80)90162-1.

20) Lim YP, Lin YC & Pandy MG. Lower-limb muscle function in healthy young and older adults across a range of walking speeds. *Gait Posture*. 2022, 94, 124-130. DOI: 10.1016/j.gaitpost.2022.03.003.

21) Tanino Y. Knowledge of Mechanics for Observational Motion Analysis. *J Kansai Phys Ther* (in Japanese). 2022, 2, 11-16.

22) McGibbon CA. Toward a better understanding of gait changes with age and disablement: neuromuscular adaptation. *Exerc Sport Sci Rev*. 2003, 31(2), 102-108. DOI: 10.1097/00003677-200304000-00009.

23) Kerrigan DC, Todd MK, Croce D, Lipsitz LA & Collins JJ. Biomechanical gait alterations independent of speed in the healthy elderly: evidence for specific limiting impairments. *Arch Phys Med Rehabil*. 1998, 79(3), 317-322. DOI: 10.1016/s0003-9993(98)90013-2.

24) Whittington B, Silder A, Heiderscheit B & Thelen DG. The contribution of passive-elastic mechanisms to lower extremity joint kinetics during human walking. *Gait Posture*. 2008, 27(4), 628-634. DOI: 10.1016/j.gaitpost.2007.08.005.