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Original Article

Influences of trunk flexion on mechanical energy flow in the lower extremities during gait

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Abstract. [Purpose] The time-series waveforms of mechanical energy generation, absorption, and transfer through the joints indicate how movements are produced and controlled. Previous studies have used these waveforms to evaluate and describe the efficiency of human movements. The purpose of this study was to examine the influence of trunk flexion on mechanical energy flow in the lower extremities during gait. [Subjects and Methods] The subjects were 8 healthy young males (mean age, 21.8 ± 1.3 years, mean height, 170.5 ± 6.8 cm, and mean weight, 60.2 ± 6.8 kg). Subjects walked at a self-selected gait speed under 2 conditions: normal gait (condition N), and gait with trunk flexion formed with a brace to simulate spinal curvature (condition TF). The data collected from initial contact to the mid-stance of gait was analyzed. [Results] There were no significant differences between the 2 conditions in the mechanical energy flow in the knee joint and negative mechanical work in the knee joint. However, the positive mechanical work of the knee joint under condition TF was significantly less than that under condition N. [Conclusion] Trunk flexion led to knee flexion in a standing posture. Thus, a strategy of moving of center of mass upward by knee extension using less mechanical energy was selected during gait in the trunk flexed posture.
Key words: Mechanical energy flow, Trunk flexion, Lower extremities

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INTRODUCTION

The time-series waveforms of mechanical energy generation, absorption, and transfer through the joints indicate how movements are produced and controlled. Previous studies have used these waveforms to evaluate and describe the efficiency of human movement[1–3)]. These waveforms can be calculated by combining joint reaction forces and joint moments with segmental and joint kinematics[4)]. Above all, it is important to investigate the work that is done with joint moments, because the energy generated by this work is used to move body segments. Williams and Cavanagh stated that mechanical energy flow is generated in intersegments, and the energy is stored in muscles, thereby reducing the generation of physiological energy by the muscle and enhancing the efficiency of the movement[5)].

Previous studies have suggested that age-related physical deformities have a variety of influences on walking efficiency[6–8)]. Kyphosis is one of the deformities in which there is alteration of trunk alignment. An epidemiological study reported that 41% of elderly people have kyphosis[9)]. A distinctive feature of kyphosis is excessive hip and knee flexion in a standing position. This compensation may be necessary to maintain the gravitational line within the base of support[10)], but increases the load on the lower extremities during standing and walking according to kinematic and kinetic studies[11–13)]. However, there is little insight into the strategy used for gait with this compensation.

To our knowledge, there are no studies that have investigated the influence of kyphosis posture on mechanical energy flow

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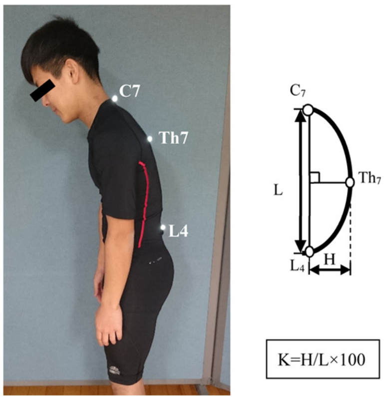

Fig. 1. The method of quantifying the degree of spinal curvature as a scale for kyphosis
C7: spinous process of seventh cervical vertebra; Th7: spinous process of seventh thoracic vertebra; L4: spinous process of fourth lumber vertebra; K: the degree of spinal curvature

in the lower extremities during gait. The influence of kyphosis on the lower extremities during gait should be investigated to prevent secondary impairments. Therefore, in this study, these influences were examined in healthy subjects as a preliminary step towards understanding the biomechanical demands on the lower extremities during the gait of elderly people with kyphosis.

SUBJECTS AND METHODS

Eight healthy young males (mean age, 21.8 ± 1.3 years, mean height, 170.5 ± 6.8 cm, and mean weight, 60.2 ± 6.8 kg) participated in our study. The subjects had no self-reported musculoskeletal and/or neurological disorders that affected their gait. This study was approved by the Institutional Review Board of Hiroshima University, and each subject was informed of the objectives of the study and provided written informed consent prior to participation.

The subjects walked along a 10 m walkway at a self-selected gait speed under 2 conditions: a normal gait, in which no brace was attached (condition N), and gait with trunk flexion formed with a brace which was attached to simulate spinal curvature (condition TF). Trials under each condition were conducted in a random order, and the measurement of each task was repeated 5 times. To quantify the degree of spinal curvature as a scale for kyphosis (K), the method of Milne and Lauder was used[14] (Fig. 1). Under condition TF, spinal curvature was simulated by attaching a brace to increase the kyphosis index to greater than 13 during standing[14].

Fourteen-millimeter diameter reflecting markers were attached to the following 36 bilateral anatomical landmarks: the temples, lateral ends of the superior nuchal line, tragus, acromion, olecranon, styloid process of the ulna, inferior edge of the last rib, great trochanter, lateral and medial epicondyles of the femur, lateral and medial condyles of the tibia, lateral and medial malleoli, first and fifth metatarsal heads, and calcaneal tuberosity. The 3-dimensional (3D) coordinates of these points were recorded using a 3D motion analysis system (Vicon motion system, Oxford, UK) with 6 cameras at a sampling rate of 100 Hz. At the same time, ground reaction force was recorded using 8 force plates (Tec Gihan, Uji, Japan) at a sampling frequency of 1,000 Hz.

The kinematic data were filtered with a cut-off frequency of 6 Hz, and the kinetic data were filtered with a frequency of 20 Hz using a fourth-order low-pass Butterworth filter. An eight-part rigid-body linked model was constructed consisting of the thorax, pelvis, thighs, shanks, and feet, using the recorded marker coordinates. Spatial coordinates were described using a right-handed coordinate system, and the medio-lateral direction was labeled the ML-axis, the anteroposterior direction as the AP-axis, and the vertical direction as the V-axis when standing. The local coordinate system was similarly defined as the absolute coordinate system with the ml-axis, ap-axis, and v-axis. The COM coordinates of the whole body, the velocities of COM, the joint and segment angle were calculated.

The initial contact of the left leg was defined as the time when the vertical ground reaction force reached 10 N. The vertical coordinates of COM were labeled COM_V and the anteroposterior component of the velocity of COM was used as the gait velocity. The stance phase of walking was defined as the time between initial contact and toe-off of the left leg, and the time between the initial contact of the left leg and the moment when COM_V reached its maximum was defined as the COM rising phase.

The means of the backward leaning angles of the thorax and thigh in the standing position, the means of gait velocity during one walking cycle and the displacement of COM_V (ΔCOM_V, the value of the maximum over the minimum), and the

Table 1. Spatiotemporal data of the two conditions

	Condition N	Condition TF
ΔCOM_V [%BH]	1.76 ± 0.27	1.48 ± 0.38*
COM-Vel$_{AP}$ [m/s]	1.15 ± 0.10	1.06 ± 0.11*
Motion time [s]	0.28 (0.27–0.31)	0.32 (0.28–0.34)
Backward leaning angles [deg]		
Thorax	5.03 ± 5.18	−28.37 ± 27.12**
Thigh	−9.29 ± 2.91	0.91 ± 7.10**
Joint moments [N·m·s/kg]		
Hip extension	0.02 (0.01–0.04)	0.14 (0.02–0.24)*
Knee extension	0.07 ± 0.02	0.08 ± 0.04

Mean ± standard deviation, * $p<0.05$, ** $p<0.01$
ΔCOM_V (the displacement of the vertical coordinates of COM) and COM-Vel$_{AP}$ (gait speed) during one walking cycle, motion time and joints moments during the COM rising phase, and backward leaning angles in the standing position were calculated

motion time during the COM rising phase were calculated the data of the markers' 3D coordinates. The joint moments were calculated using the data of the marker's 3D coordinates and the ground reaction force data during the COM rising phase by integrating the joints moments with respect to time. The rate of work acting on the joint (power) by the joint moment (i.e., joint power) was calculated as follows:

$$P(\mathrm{j}) = M(\mathrm{j})\cdot\omega(\mathrm{j}) \quad \text{(Eq. A)}$$

where P(j) is the mechanical power acting on the joint, M(j) is the joint moment acting on the joint, and ω(j) is the angular velocity of the joint. Similarly, the power generated by the joint moment acting on the segment at the joint was calculated as follows:

$$P(\mathrm{s}) = M(\mathrm{s})\cdot\omega(\mathrm{s}) \quad \text{(Eq. B)}$$

where P(s) is the mechanical power transferred to or from segment at its joint, M(s) is the joint moment acting on the segment at the joint and ω(s) is the angular velocity of the segment. A positive power indicates mechanical energy generated by the muscle of the joint, while a negative power indicates mechanical energy absorbed by the muscle of the joint[15–17]. Additionally, in terms of joint power, a positive or a negative value indicates whether the muscles of a joint were contracting concentrically or eccentrically, respectively. The mechanical work done by the joint moment during the COM rising phase was calculated by integrating the power at the joints and at the endpoints of adjoining segments with respect to time. The positional relationship with the pelvis was used to determine the expression of the proximal or distal portions in segments.

Statistical analyses were performed using statistical software SPSS 22.0 for Windows (IBM, Tokyo, Japan). The Shapiro-Wilk test was used to test the normality of the data. The paired t-test or Wilcoxon Signed Rank tests in cases of non-normality, to test statistical significance of differences between the conditions. Significance was accepted for values of $p<0.05$.

RESULTS

The thorax backward leaning angle under condition TF was significantly lower than under condition N, and the angle of the thigh under condition TF was significantly higher than under condition N during standing ($p<0.01$, Table 1). ΔCOM_V during the walking cycle under condition TF was significantly lower than that under condition N ($p<0.05$). The mean of gait velocity during the walking cycle under condition TF was significantly lower than that under condition N ($p<0.05$). There was no significant difference in the motion time of the COM rising phase between the two conditions (Table 1). The moment of the hip extension during the COM rising phase under condition TF was significantly higher than under condition N ($p<0.05$). There were no significant differences in the moments of the knee extension during the COM rising phase between the two conditions (Table 1). The positive mechanical work in the proximal portion of the thigh during the COM rising phase under condition TF was significantly higher than that under condition N ($p<0.05$). The positive mechanical work of the knee joint during the COM rising phase under condition TF was significantly lower than that under condition N ($p<0.05$). There were no significant differences in the positive mechanical work in the distal portion of the thigh, the negative mechanical work in the proximal portion of the shank, or the negative mechanical work of the knee joint of the COM rising phase between the two conditions (Table 2, Fig. 2).

Table 2. Mechanical work performed under the two conditions

	Condition N	Condition TF
Mechanical works [J/kg]		
Thigh $_{proximal}$ *Pp*	0.02 (0.01–0.05)	0.13 (0.02–0.21)*
Thigh $_{distal}$ *Pp*	0.08 (0.05–0.13)	0.09 (0.04–0.15)
Shank $_{proximal}$ *Np*	0.10 ± 0.05	0.13 ± 0.09
Knee *Pp*	0.02 ± 0.01	0.01 ± 0.01*
Knee *Np*	0.04 ± 0.02	0.06 ± 0.03

Mean ± standard deviation or median (IQR), * $p<0.05$.
Thigh $_{proximal}$ *Pp*: Positive power in the proximal portion of the thigh
Thigh $_{distal}$ *Pp*: Positive power in the distal portion of the thigh
Shank $_{proximal}$ *Np*: Negative power in the proximal portion of the shank
Knee *Pp*: Positive power in the knee joint
Knee *Np*: Negative power in the knee joint

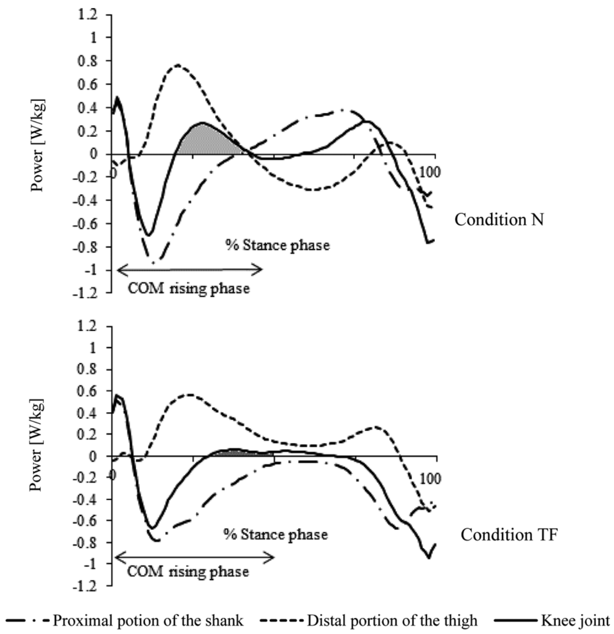

Fig. 2. The time-series waveforms of mechanical energy in the knee during gait
Lines represent mean powers normalized to body mass (W/kg). No significant difference was seen in the mechanical energy flow of the knee between the two conditions (the power of the proximal portion of the shank and the distal portion of the thigh did not change); however, the positive mechanical work of the knee joint under condition TF was significantly lower than that under condition N during the COM rising phase (shaded area).

DISCUSSION

A human efficiently moves the COM forward and walks by acquiring potential energy during the COM rising phase and converting it into kinetic energy[18, 19]. In this study, ΔCOM_V and the mean of gait velocity were significantly lower under condition TF than those under condition N. Potential and kinetic energies are proportional to the position and square of velocity, respectively; thus, our results show that the amount of potential energy that can be converted to kinetic energy was reduced under condition TF, and, as a result, the gait speed was also reduced. To determine the cause, change in the activity of the lower extremities was investigated focusing on the phase when COM was moving upward and forward.

Here, how mechanical energy flow was generated in the knee joint under condition N during the COM rising phase is discussed. The power showed a negative value in the proximal portion of the shank and a positive value in the distal portion of the thigh, and there was a transition from a negative to positive value at the knee joint. The knee extensor muscles absorbed mechanical energy as a result of the forward leaning movement of the shank, and then they contracted concentrically. Eventually, as mechanical energy was generated, the knee extensor muscles caused forward leaning movement of the thigh. Thus, knee extension was performed efficiently using mechanical energy during this phase and utilization of physiological energy by the muscles was suppressed under condition N[5]. Previous studies have demonstrated that humans perform knee extension most efficiently during the phase from the loading response to mid-stance by contracting the quadriceps first eccentrically and then concentrically, similar to a spring[19, 20]. Thus, under condition N, the movement was performed efficiently using mechanical energy and moved COM upward during the COM rising phase.

The power showed a negative value in the proximal portion of the shank and a positive value in the distal portion of the thigh under condition TF, and no significant difference found between the two conditions in the mechanical work performed by either of them, or the negative work of the knee joint. However, the positive mechanical work of the joint under condition TF was significantly lower than that under condition N. These results indicate there was little difference in mechanical energy flow between the two conditions, but the amount of work done by knee extension under condition TF was significantly lower than that under condition N. Stretching the lower extremities by extending the knee moves COM upward during the COM rising phase. A previous study reported that the rate of utilization of gravity during walking is related to the knee extension angle and the positive work of the knee joint during the COM rising phase[21]. Thus, ΔCOM_V would decrease because knee extension could not be performed sufficiently to move COM higher.

Regarding the hip joint during the COM rising phase, the hip extension moment and the positive mechanical work in the proximal portion of the thigh under condition TF were significantly higher than those under condition N. These results indicate that modifying the demands of the muscular control of the hip extensor muscles increased the mechanical energy generated by the proximal portion of the thigh under condition TF. Previous research has shown that standing or walking with the trunk flexed increases the muscular demands on the hip extensor muscles to maintain the posture[22]. Therefore, our results may show that the demands on the hip extensors were increased by sustained hip flexion because the knee could not be sufficiently extended.

Consequently, there were no significant differences in mechanical energy flow in the knee between the two conditions during the COM rising phase. The results of this study also suggest that a strategy with less mechanical energy was chosen, because the knee could not be sufficiently extended, and upward movement of COM decreased in the trunk flexed posture. Finally, the mechanical demand on the hip extensor muscles may increase because the knee cannot be extended sufficiently, and the demands may cause secondary disabilities.

This study had some limitations. First, the sample size was small and the subjects were healthy young people; therefore results cannot be generalized to elderly people with kyphosis. Further studies of elderly people with kyphosis are needed. Second, a brace was used to set the degree of trunk flexion. Therefore, it is unclear whether the exact same conditions were replicated for each subject. However, by observing the tendencies in our data the general influence of the posture on the lower extremities was determined. Third, the kinetics data provide only limited insight into muscle activities using an inverse dynamics approach. Finally, the values of mechanical energy were interpreted without observing the joint movement in detail, relying on the values and waveforms of the power and joint moments. Therefore, it will also be necessary to determine the relationship between the joint movement and the kinetic aspects.

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