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Original Article

Estimation of the ankle power during the terminal stance of gait using an inertial sensor

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Abstract. [Purpose] The purpose of this study was to develop an assessment tool that reflects the ankle function during the terminal stance of gait using an inertial sensor. [Participants and Methods] Thirteen healthy males (20 limbs) participated in this study. All the participants were required to perform five straight-line walking trials along a 10-m level walkway. During the terminal stance phase, both the anterior-posterior and vertical accelerations were measured with an inertial sensor mounted on the fibular head. The Pythagorean theorem was used to calculate the acceleration vector. A three-dimensional gait analysis system was used for movement data acquisition. All statistical analyses were performed using IBM SPSS Statistics 24.0 for Windows. [Results] Results were obtained using the following multiple regression equation for the estimation of ankle plantar flexion power: Estimated Ankle Power=−4.689 + 0.269 × vertical acceleration + 0.104 × body weight. [Conclusion] Our novel method for gait analysis using an inertial sensor can assess the ankle power during the terminal stance phase of gait.

Key words: Inertial sensor, Gait analysis, Ankle plantar flexion power

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INTRODUCTION

Physical therapy is performed at rehabilitation centers as well as in the home and community of patients. In particular, many elderly performed physical therapy at home and community of the patients. In order to maintain quality of life (QOL) and Activity of Daily Living (ADL), it has been reported that walking speed is important[1, 2)]. Therefore, in order to improve QOL and ADL, it is important that physical therapists to assess gait performance. However, it is unable to quantitatively assess the gait ability in the home and community of the patients.

In general, spatio-temporal analyses of walking in the elderly demonstrate a slower walking speed, smaller step length[3)], and an increased double-support stance period[4)] compared with the walking of younger persons. Winter reported kinetic data indicating that push-off power during the terminal stance phase of the elderly was decreased compared with that of young persons[4)]. Judge et al.[5)] reported that the elderly showed decreased peak plantar flexor moment and plantar flexor power in the terminal stance during gait compared with younger persons, while peak hip and knee extensor power were similar between elderly and young persons. These gait kinetics indicate that the elderly may compensate for decreasing peak ankle plantar flexor power by increasing hip flexor power[5)]. Moreover, they suggest that maintaining step length is important in advanced age for ankle plantar flexor muscle training[6)]. Thus, it is evident from these findings that walking of elderly is decrease ankle plantar flexor power from early stage.

In the clinical setting, gait observation is important in determining the effect of physical therapy treatment. In some studies,

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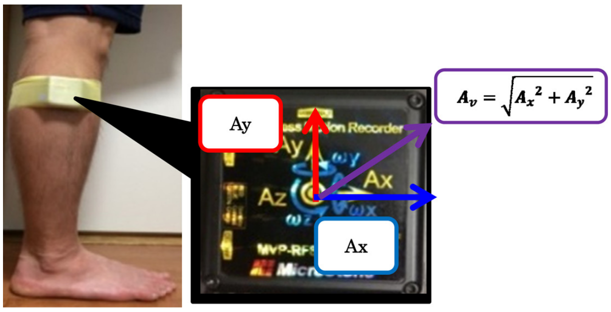

Fig. 1. An inertial sensor mounted on the lower leg at the fibular head.
The inertial sensor recorded the direction of acceleration of the proximal shank during gait: anterior–posterior, vertical, and medial–lateral. During the terminal stance phase, both the anterior–posterior acceleration (Ax) and vertical acceleration (Ay) were measured by the inertial sensor mounted on the fibular head.
The Pythagorean theorem was used to calculate the acceleration vector (Av) as follows: $Av = \sqrt{Ax^2 + Ay^2}$.

the reliability of observational gait analysis shows moderate inter-observer reliability and moderate or good intra-observer reliability[7, 8]. Conversely, observers recorded only 22.2% of the predicted gait deviations in another study[9]. Opinion varies as to reliability in gait observation. In order to create reliable observational analysis, it is necessary to development more reliable devices for analysis. The use of video reportedly improves the reliability of observation analysis[10]. Further, the causes of gait disorder must be examined.

An inertial sensor is typically lightweight, portable, easy to use, and does not require a special environment. These characteristics allow for quantitative analysis of walking in the clinical practice. Therefore, inertial sensor-based activity monitors have become popular recently, and can be used to evaluate the gait of patients during rehabilitation. Inertial sensors were recognized as validated, objective tools for quantitative assessment without the need for specific environments[11, 12]. Inertial sensors could measure the center of gravity if the sensor is fixed to the pelvis[8]. Although studies have been made on movement fluency and regularity quantitatively, there is little agreement on theory exists for analyzing movement disorders. However, the analysis methods had begun to change since around 2008. Rencheng reported three motion sensor units mounted on the foot, calf and thigh estimated joint position and joint moment and power[13]. Rouhani studied an ambulatory system consisting of plantar pressure insole and inertial sensor on foot and shank was used, and the proposed ambulatory system could be easily assess main ankle kinetics for clinical applications[14]. There is possibility that an inertial sensor can be analysis kinetics from previous studies. Therefore, the purpose of this study was to develop an assessment that reflects ankle function during the terminal stance of gait using a single inertial sensor mounted on the fibular head.

PARTICIPANTS AND METHODS

Twenty limbs of 13 healthy male (mean age, 24.3 ± 5.5 years; mean height, 170 ± 4.1 cm; mean weight, 60.7 ± 3.7 kg) were included in this study. Participants whose kinematic parameters fell outside of the normal ranges were excluded. This study was approved by the Morinomiya University of Medical Science ethics committee (Approval number: 2016-03), and all participants provided informed consent. All participants were required to perform 5 straight-line walking trials along a 10-m level walkway. Step length was calculated by height at 3 rhythms: slow (76 steps/min), middle (108 steps/min), and fast (125 steps/min). Participants were equipped with an inertial sensor mounted on the fibular head. The inertial sensor recorded the direction of acceleration of the proximal shank during gait: anterior–posterior, vertical, and medial–lateral. During the terminal stance phase, both the anterior–posterior acceleration (Ax) and vertical acceleration (Ay) were measured by the inertial sensor mounted on the fibular head (Fig. 1). The Pythagorean theorem was used to calculate the acceleration vector (Av) as follows: $Av = \sqrt{Ax^2 + Ay^2}$ [15].

Gait cycle was examined using video image files captured by a tablet PC (motion recorder MYP-RF8-TS; sampling frequency, 100 Hz; MicroStone Corporation) synchronized with the inertial sensor. Both Ax and Ay peaked from heel-off (HO) to toe-off (TO) (Fig. 2).

A three-dimensional (3D) gait analysis system was used for movement data acquisition. The system consisted of 6 infrared cameras (sampling frequency, 100 Hz; VICON) and 2 force plates (sampling frequency, 1,000 Hz; AMTI). Thirty-five reflective markers were mounted on the skin using double-sided adhesive tape following the plug-in gait model. Kinematic lower limb data were calculated using Nexus (VICON), and a 10-Hz low-pass filter was applied. Ankle plantar, knee flexor, and hip flexor peak moments, as well as peak ankle plantar flexion power, the force time integral of ankle power (negative, loading response to terminal stance; positive, terminal stance to initial swing), and hip, knee, and ankle angles were calculated from the Nexus data. Moreover, to calculate the synthetic vector (Av), the following formula was used: $Av = \sqrt{Ax^2 + Ay^2}$.

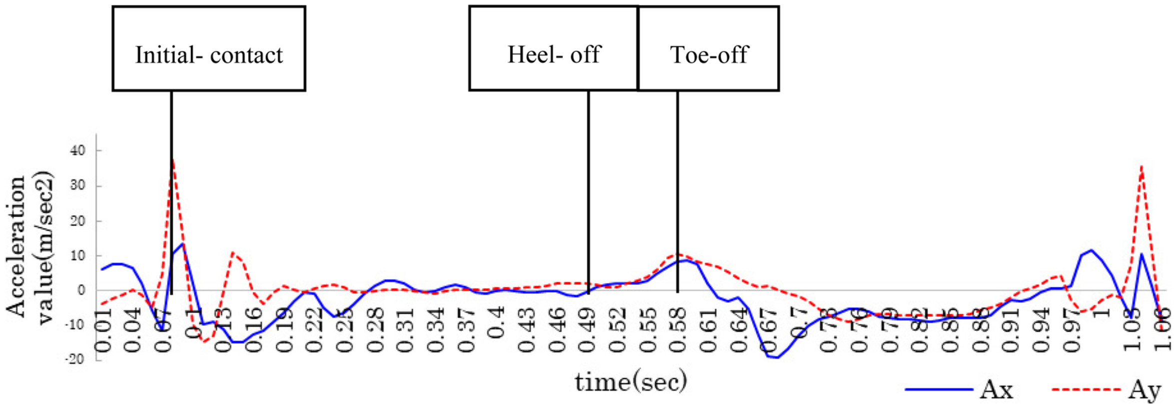

Fig. 2. Acceleration of the Fibular head during gait (gait condition: fast).
Ax: anterio-posterior direction; Ay: vertical direction.
The peak values of both anterior-posterior direction (solid blue line) and vertical direction (dotted red line) from Heel-off to Toe-off were measured.

All parameters were assessed among the 3 conditions (slow, middle, and fast walking speed) using a repeated analysis of variance (ANOVA). Relationship acceleration parameters as both Ax, Av, and ankle kinetics during gait were assessed using Spearman's correlation coefficient. Multiple regression analysis was carried out using a stepwise method to determine correlations with the Av. The Av was considered a dependent variable. Data from hip extension, knee flexion, ankle plantar flexion angles, peak ankle plantar flexion power, positive vertical impulse, negative vertical impulse, peak ankle plantar flexion moment, peak hip flexion moment, peak knee flexion moment, and support moment items were considered independent variables. Finally, multiple regression analysis was carried out using a stepwise method to obtain the estimated ankle power.

All statistical analyses were performed using IBM SPSS Statistics 24.0 for Windows. Values of $p<0.05$ were considered to indicate statistical significance for all tests.

RESULTS

Using our methods, 100% of HO to TO events were detected. When the walking speed increased, the peak ankle plantar flexion moment, peak ankle plantar flexion power, Ax, and Av were increased significantly over that recorded at slower speeds. However, ankle plantar flexion angle was not significantly different among three speed conditions of walking (Table 1).

The repeated measures ANOVA revealed significant differences among the 3 walking speed conditions for all variables except the ankle plantar flexion angle and peak knee flexion moment ($p<0.05$; Table 1).

There was a significantly weak correlation between Ax and knee flexion angles ($r=0.35$, $p<0.05$; Table 2) and Av and knee flexion angles ($r=0.35$, $p<0.05$; Table 2). Peak ankle plantar flexion power was correlated significantly and strongly with Ax and Av, respectively (Table 2). Therefore, the inertial sensor mounted on the fibular head allowed for assessment of ankle function, which is important for forward progression force during the terminal stance phase.

The results of the stepwise multiple regression analysis is shown in Table 3. This model accounted for 70.8% of the variation in the acceleration parameter as Av among the variables (Adjusted $R^2=0.687$).

Multiple regression analysis was performed using a stepwise method to obtain the estimated ankle power. The ankle power was considered a dependent variable. Data from age, height, body weight, and acceleration parameters (Ax, Ay, Av) items were considered independent variables. Results were obtained in the following equation for the estimation of ankle plantar flexion power (Table 4; Adjusted $R^2=0.54$).

Estimated Ankle Power (W)= $-4.689 + 0.269 \times \text{Ay} + 0.104 \times \text{Body Weight}$

DISCUSSION

We hypothesized that gait analysis using an inertial sensor could detect ankle power during the terminal stance phase of gait. This study examined lower leg acceleration during the terminal stance phase to extract an acceleration parameter closely correlated to kinematic and kinetic variables. Our findings indicate that an inertial sensor mounted on the fibular head can assess ankle function, which is important for forward progression force during the terminal stance phase.

In this study, the walking speed changed under different walking conditions, while the step length remained constant. As a result, when the walking speed increased, the peak ankle plantar flexion moment, peak ankle plantar flexion power, and acceleration parameters increased, with a significant difference between the three walking speed groups. Therefore, kinematic

Table 1. Summary of test results, mean (standard deviation) value of participant

	Variables	SLOW	FREE	FAST	Significant difference
Moment (N/mm)	Ankle plantar flexion	138.4 (20.8)**	148.7 (22.6)**	157.4 (23.5)**	SLOW<FREE, FREE< FAST, SLOW<FAST
	Knee extension	21.6 (11.3)	29.6 (11.9)	35.1 (11.3)	-
	Hip flexor	−82.0 (17.7)**	−107.4 (20.4)**	−125.6 (21.1)**	SLOW<FREE, FREE< FAST, SLOW<FAST
	Support	78.0 (30.9)**	70.9 (30.3)**	66.8 (31.8)**	SLOW<FREE, FREE< FAST, SLOW<FAST
Power (w)	Ankle plantar flexion	2.54 (0.67)**	4.05 (0.87)**	5.03 (0.95)**	SLOW<FREE, FREE< FAST, SLOW<FAST
Vertical impulse (w)	Positive	0.29 (0.08)**	0.33 (0.08)**	0.36 (0.09)**	SLOW<FREE, FREE< FAST, SLOW<FAST
	Negative	0.18 (0.07)**	0.15 (0.07)**	0.14 (0.06)**	SLOW<FREE, FREE< FAST, SLOW<FAST
Angle (degrees)	Ankle plantar flexion	17.4 (5.8)	19.3 (4.7)	19.1 (5.6)	-
	Knee flexion	56.0 (5.3)**	61.5 (5.3)**	62.5 (5.4)**	SLOW<FREE, FREE< FAST, SLOW<FAST
	Hip extension	16.0 (6.4)**	17.0 (6.7)**	17.8 (6.9)**	SLOW<FREE, FREE< FAST, SLOW<FAST
Acceleration parameters	Ax(m/sec^2)	5.9 (1.6)**	10.3 (2.5)**	13.1 (3.3)**	SLOW<FREE, FREE< FAST, SLOW<FAST
	Av	6.6 (2.08)**	12.2 (2.5)**	15.7 (3.4)**	SLOW<FREE, FREE< FAST, SLOW<FAST

**$p<0.05$, *$p<0.01$.

Table 2. Variables that correlate with the Ax and Av

	Moment				Power	Vertical impulse		Angle		
	Ankle plantar flexion	Knee flexion	Hip flexor	Support	Ankle plantar flexion	Positive	Negative	Ankle plantar flexion	Knee flexion	Hip extension
Ax	0.24	0.37*	0.62*	−0.24	0.64**	0.22	−0.20	0.16	0.35*	0.19
Av	0.30**	0.37*	0.64*	−0.20	0.71**	0.26	−0.21	0.10	0.33*	0.24

**$p<0.05$, *$p<0.01$.

Table 3. Results of multiple regression analysis (The dependent variables: Av)

	Unstandardized Coefficients	Standardized Coefficients β	95% Confidence Interval Lower Bound	95% Confidence Interval Upper Bound	p-value	VIF
(Constant)	0.856		−3.694	5.407	0.708	
Ankle plantar flexion power	3.353	0.953	2.396	4.309	0.000	3.469
Hip flexor moment	−0.005	−0.279	−0.008	−0.002	0.003	1.484
Ankle plantar flexion	0.155	0.181	0.024	0.287	0.021	1.096

Adjusted R^2=0.687.

Table 4. Results of multiple regression analysis (The dependent variables: Ankle plantar flexion power)

	Unstandardized Coefficients	Standardized Coefficients β	95% Confidence Interval Lower Bound	95% Confidence Interval Upper Bound	p-value	VIF
(Constant)	−4.689		−8.749	−0.629	0.024	
Ay	0.269	0.680	0.199	0.340	0.000	1.001
Body weight	0.104	0.278	0.038	0.170	0.003	1.001

Adjusted R^2=0.536.

variables, kinetic variables, and acceleration parameters are changed by the changing the walking speed. Moreover, after univariate analyses, Multiple regression analysis was performed using a stepwise method to calculate the estimated ankle plantar flexion power by the acceleration parameter. Consequently, vertical acceleration parameter as Ay and body mass were chosen as variables related to ankle plantar flexion power.

Kavanagh reported that the reliability of an inertial sensor during clinical gait analysis was high. Specifically, the interclass correlation coefficient (ICC) of the of the shank segmental accelerations during gait was 0.94[12]. A previous gait analysis study used an inertial sensor to examine walking velocity (m/s), cadence (steps/min), average step length (cm), step timing variability, acceleration root mean square (RMS), and the harmonic ratio of acceleration signals[16]. Most gait analyses using an inertial sensor have described physical activity objectively[17, 18] ; assessed the walking stability of patients with diabetic peripheral neuropathy[19], and analyzed patients with osteoarthritis during gait[20–22].

In previous research, inertial sensors were used to assess quantitatively the fluency and regularity of gait. However, Turcot showed that an inertial sensor assessed the lateral thrust magnitude of the osteoarthritic knee during the gait[23]. Gait analysis using an inertial sensor can demonstrate gait performance, however, very few attempts have been made at assess the kinetics of gait. In this study, we identified a novel method to estimate the ankle joint power using an inertial sensor mounted on the lower leg during HO to TO.

We hypothesized that kinetics of the ankle during gait can provide gait analysis using an inertial sensor. Ankle power is product of the ankle plantar flexor moment and angular velocity. Ankle moment is expressed as a product of the ankle force and lever arm as the ground reaction force vector to the center of the ankle joint. Force is the product of mass and acceleration (Newton's equation of motion). In this study, the foot and lower leg mass is necessary to estimated ankle power. The mass of both the foot and leg are estimated as 5–6.8% of body mass[24]. Recently, Thompson reported[24] that effective mass provides an important link between vertical impact forces and tibial acceleration. In other words, body weight for estimating lower leg mass and ground force reaction vertical vectors were estimated by the lower leg vertical direction acceleration parameter. Therefore, there was shown validity in this study result from these references. In this study, vertical acceleration parameter as Ay and body mass was extracted to related variables with ankle power from multiple regression analysis. Therefore, estimated ankle power can be estimated by measurement of the acceleration generated in the lower leg and body weight.

Our method may be applied to the risk test of locomotive syndrome. Locomotive syndrome (LS) is defined as a loss of motor function as a result of disorders of motor organs[25]. In particular, Of the LS risk test, two-step test can be evaluated walking ability[26]. However, two-step test cannot be assess accurately walking ability of locomotive syndrome. Our novel method will be useful to identify deteriorating function of the ankle during the terminal stance phase of gait and allow. Thus, a device to assess ankle function quantitatively during the terminal stance phase of gait is important for early detection of LS.

Our study has limitations. First, this study was a laboratory-based gait analysis study. Function of the inertial sensor may depend on the external environment such as the road surface, the road slope, and personal shoes. Therefore, studies using outdoor-based measurements are necessary. Second, the analysis is difficult because abnormal kinematics such as limited range of motion of the lower limb joints may not provide accurate acceleration wave forms during the gait.

Future studies should examine whether this quantitative method can be used in the elderly and individuals with gait disorder. We developed a novel method for gait analysis using an inertial sensor to assess function of the ankle during terminal stance phase of gait.

Conflict of interest

We have nothing to declare for this study.

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