A Method for Gait Analysis in a Daily Living Environment by Body-Mounted Instruments*

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A new method is proposed to investigate kinematics and dynamics of locomotion without any limitation of laboratorial conditions. This method enables simple measurements and numerical kinematical calculations with small body-mounted instruments utilizing accelerometers and gyroscopes. Temporal parameters of gait are determined by classification of accelerations and angular velocities. Joint angles, joint moments, and energy consumptions as mechanical works are estimated assuming rigid-body dynamics in sagittal plane. The method is verified by comparing the results of video-based motion capture system and force plates as conventional standards, evaluating accuracy for normal level walking at four cadences. No significant differences are observed between the results of the proposed method and those from the conventional method. It confirms the validity of the proposed method for healthy subjects and the method can provides free mobility in the investigation of locomotive dynamics beyond the gait laboratory.

***Key Words*:** Information Processing and Signal Analysis, Biomechanics, Gait Analysis, Accelerometer, Gyroscope

1. Introduction

Falls due to locomotive impairment cause a serious hazard to elders, which has lately become a public interest[1]. Some reports show one-third of the elders living at home fell down each year[2]. In order to prevent these falls, we need to take a positive approach to identify their personal risk of falls in the same environments and situations in which people actually live. However, this is not still achieved because of insufficient instrument and analytical method in daily living environments.

Several quantitative methods have been proposed to assess normal and pathological gait in laboratorial environments, i.e. video-based motion capture systems, force plates, mechanical goniometers, and electromagnetic tracking systems. These are excellent methods, but unfortunately, they require considerable special floor space, exaggerate equipments, restrictive attachments or patients coming to the laboratory. On the other hand, some portable instruments for continuous gait assessments are proposed. Foot pressure distribution acquisition systems are provided to measure temporal parameters or ground reaction forces[3]. But it still have problem of high non-linearity, hysterisis, and temperature dependence of electric capacitive sensors. An analysis based on accelerometry has proved to be an excellent alternative method to investigate individual characteristics of locomotion in daily living environments[4]. Recently, technical progress has made it possible to realize small and low power consumptive accelerometers, as a promising tool for long-term ambulatory monitoring. Ambulatory monitoring systems utilizing accelerometers have been applied for the assessment of physical activity[5] or simple classification of human behaviors[6],[7]. These are based on sort of relationships between accelerometer output and metabolic energy consumptions. Some accelerometric algorithms have also been proposed to assess precise temporal gait characteristics[8]. Aminian et al. presented an

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algorithm for the detection of temporal gait parameters, such as gait cycle, stance and swing phase, using two accelerometers attached on the thighs of both legs[9]. These methods present quantitative and satisfactory evaluation, but insufficient to evaluate subsequent gait improvements with exact interpretations of dynamics of locomotion.

To overcome these limitations, an alternative approach is required that can investigate kinematics and dynamics of locomotion without any limitation of environmental conditions. One possible approach is to capture the gait kinematics utilizing completely body-mounted instruments. It requires a smart sensing device that is small, lightweight, robust, and low restrictive. A few studies have been reported on measurement techniques utilizing accelerometers or gyroscopes to investigate kinematics of human behaviors[10]-[12]. Willemsen et al. presented an advanced method for direct measurement of temporal parameters of gait and joint angles using five pair of one-dimensional accelerometers for each extremity, as an artificial feedback system of functional neuromuscular stimulation[13]-[15]. Unfortunately, the method is rather complicated due to large number of sensor attachments, and not proper for further investigation of locomotive dynamics.

This study is performed for total and quantitative gait assessment in the field studies beyond the laboratory environments. An advanced method is proposed that can achieve continuous kinematical analysis by using low restrictive body-mounted instruments. Detection of temporal gait parameters, calculation of joint angles, and estimation of energy consumptions as mechanical works via inverse dynamic calculations are presented.

2. Method

We have developed body-mounted instruments to measure dynamics of locomotion. For simplification, a two-dimensional model of leg motion is considered assuming that it is sufficient to represent biped locomotion in sagittal plane. Additionally the method is designed to be functional against single leg motion. Thus when we can assume bilateral symmetry of gait, we can eliminate sensor attachment for the other leg, and contribute to reduce power consumption and body restriction.

A sensor unit consists of a pair of one-dimensional accelerometers (Analog Devices ADXL05) and one gyroscope (Murata ENC-03J). Total size of the unit is 25×33×10 mm. The accelerometer is a DC accelerometer having range of ±5 G, which can measure not only accelerations caused by force imposed on a segment but also static accelerations

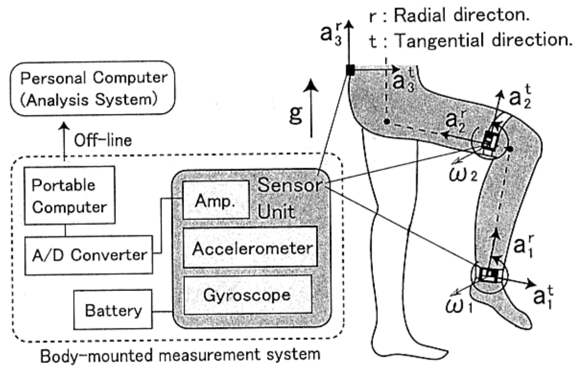


Fig. 1 Schematics of the completely body-mounted instruments. A battery and connectors are mounted at the belt. Portable A5 size notebook computer is carried with a backpack.

such as gravitational acceleration. These characteristics are suitable for postural calculation as a tilt reference. The gyroscope has small and quick response up to 50 Hz, and wide range of ±300 deg/sec. The signals are amplified, sampled at 100 Hz via PCMCIA card type 12-bit A/D converter, and recorded to the compact A5 size notebook computer.

Measurements were made with six normal adults free of any disease. All subjects signed consent forms. The sensor units were attached to distal position of shank, thigh of right leg, and center of lumber with Velcro straps (Fig. 1), assuming bilateral symmetry of normal gait. The notebook computer was carried with a backpack. Walking speed was controlled by cadence at 60, 80, 100, 120 step/min. Usual walking cadence of an adult is about 100 step/min. Subjects were instructed to walk straight along 6 m normally adjusting theirselves to metronome beep at their own preferred step lengths. To evaluate the accuracy of the proposed method, three-dimensional postural data and ground reaction forces were also collected by a video-based motion capture system (Oxford Metrics Vicon512) and six force plates (AMTI OR6) positioned along the walkway, as a gold standard of conventional method. Prior to data recording, optical markers were placed at both sides of shoulder, pelvis, hip joints, knee joints, ankle joints, fifth metatarsal and heel of feet. Each recording session was synchronized and the data in the notebook computer were transferred to a computer for analysis at the end of all recording sessions.

3. Algorithm

3.1 Detection of temporal parameters

The outputs of accelerometers attached on the human body include several frequency components. The higher frequencies are caused by the impact of heel contact, and the lower frequencies are directly result from voluntary kinetic body movements or inclinations with respect to the gravity direction (see, Fig. 2). It suggests that characteristics of frequency components of acceleration can be used as detectors of the temporal parameters. In our experiments, most of the power spectrum of low frequency acceleration is concentrated below 5 Hz, and power spectrum of high frequency is concentrated above 30 Hz in normal walking. Consequently, digital 4th-order Butterworth low-pass and high-pass filters, cutoff frequencies of 10 and 30 Hz respectively, are designed to get optimized signals for the detection of temporal parameters. The detection process is applied to the radial acceleration measured at shank, which shows significant variations with major changes of foot kinematics during locomotion.

Foundation of the algorithm is the recursive detection of conditions given in Eqs. (1) - (5) respectively (see, Fig. 3). Distinct spikes in high frequency component are recognized as heel contact (H.C.) events. Point A is detected as a peak of the spike. These kinds of spikes are sometimes caused by foot release and are main source of false detection. To prevent false detection of the heel contact, the algorithm refers to the signal of the angular velocity ω_2

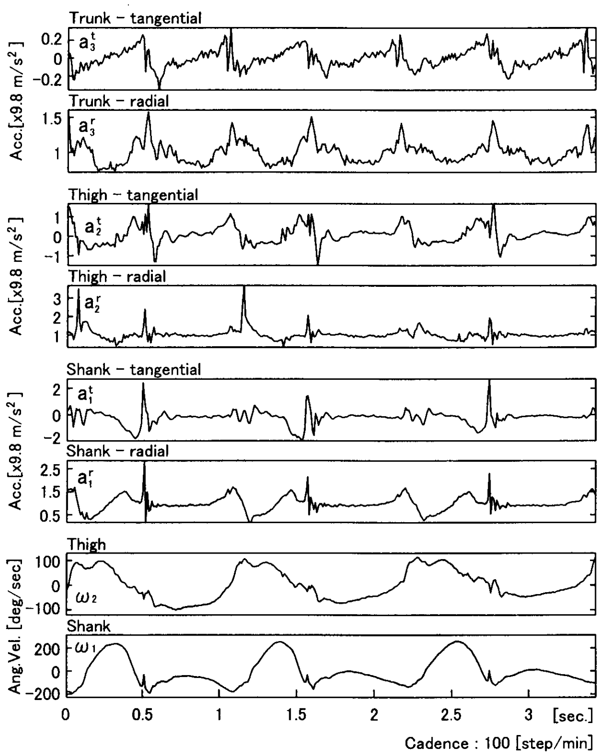


Fig. 2 A typical example of accelerations and angular velocities as measured at cadence 100 step/min of normal level walking. Frame configurations of each component are shown in Fig. 1. Characteristic waveforms are observed with ground contact event.

measured at the distal position of thigh. The derivatives of ω_2 is always negative around the true heel contact instance.

$$\left\{ t_n^A \left| \begin{array}{l} t_n = j|\delta t| + \{t'_{j,k}| \max(|\alpha_1^r(\delta t)|)\}, \\ \alpha_1^t(t_n) > 2\sigma(|\alpha_1^t|), \\ \omega_2(t_n + |\delta t|) - \omega_2(t_n - |\delta t|) < 0, \\ \omega_2(t_n + |\delta t|) \cdot \omega_2(t_n - |\delta t|) < 0 \end{array} \right. \right\} \quad (1)$$

$$n = 1, 2, \cdots, N, \ j = \{0, 1, \cdots, [t_N/\delta t]\}$$
$$\delta t = [t'_{j,1}, \cdots, t'_{j,20}], \ |\delta t| = 0.2s.$$

Where, t_n^A represents time progress to the nth event of A. The notation $|\alpha_1^r|$ represents absolute value of the high frequency component of the radial acceleration at shank. δt is a short-term calculation window for recursive algorithm. The width of calculation window $|\delta t|$ is constant. Suffix j is discrete number for each calculation window, and k is the discrete number of serial data from the beginning of δt. The notation $t'_{j,k}$ represents time progress in δt. Considering normal gait, two phases are involved in single support phase namely, mid stance and terminal stance[16]. Mid stance is in the first part of single support phase of right/left leg that begins as left/right foot is lifted and continues until heel off (H.O.) instance. Terminal stance begins with right/left heel rise and continues until left/right foot strikes the ground. In mid stance, the foot of supporting leg is stationary and the acceleration measured at shank is considered constant. Points B and C can be considered as the beginning of single support phase (SSPb) and H.O. instance respec-

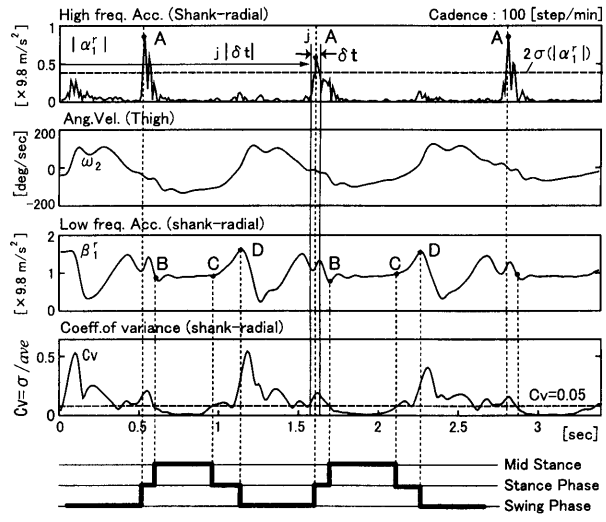


Fig. 3 The set of signals processed for the temporal parameter determination. The points A, B, C, D are detected as candidate points for H.C., SSPb, H.O., T.O. respectively. The detection algorithm is applied recursively with the calculation window δt. The diagram at the bottom of the figure indicates phase of gait determined by the point A, B, C, D.

tively. The points B and C are detected by the level of coefficient of variance Cv as follows.

$$\left\{t_n^B \middle| \begin{array}{l} t_n=j|\delta t|+\{t'_{j,k}|Cv(t'_{j,k})<0.05\},\\ \min\{t_n|t_n>t_n^A\}\end{array}\right\} \quad (2)$$

$$\left\{t_n^C \middle| \begin{array}{l} t_n=j|\delta t|+\{t'_{j,k}|Cv(t'_{j,k})>0.05\},\\ \min\{t_n|t_n>t_n^B\}\end{array}\right\} \quad (3)$$

$$Cv(t'_{j,k})=\sigma\{\beta_1^r(\delta t)\}/\mathrm{mean}\{\beta_1^r(\delta t)\}. \quad (4)$$

Where, β_1^r represents low frequency component of the radial acceleration at shank. σ denotes standard deviation. The point D as toe off (T.O.) event is detected as.

$$\left\{t_n^D \middle| \begin{array}{l} t_n=j|\delta t|+\{t'_{j,k}|\max(\beta_1^r(\delta t))\},\\ \min\{t_n|t_n>t_n^C\}\end{array}\right\}. \quad (5)$$

3.2 Body model for kinematics

Human body is represented by the five segments rigid-body linked model in two dimensions consisting of two segments for each lower extremity and one for the head, arms, and trunk (HAT). The inertial contribution by foot to the ankle moment is neglected. The model has ideal pin joints without damping factor. Trunk is only allowed to perform vertical and horizontal motion without rotation of the body. The use of the five segments model requires these assumptions, however, it is sufficient to investigate fundamental kinetics and mechanical characteristics of locomotion. The link and body parameters such as mass of the segment, position of the center of the mass, and radial of gyration of the segment are estimated using published anthropometrical data[17].

3.3 Calculation of joint angles

Joint angles are calculated by the integration of angular velocities with automatic resetting of integrators and drift corrections.

$$\theta_{i,n}=\int_{\acute{T}_n}(\omega_i-\omega_{i-1})dt+\theta_{i,n}^{ini},\quad \omega_0=0. \quad (6)$$

Joint angle $\theta_{i,n}$ is defined as the relative angle between body fixed frame $\{i\}$ and frame $\{i-1\}$ for nth stride. Integration time $\acute{T}_n$ is equal to the duration from P_n to P_{n+1} (see, Fig. 4). P_n is the peak point of angular velocity of shank in mid stance. The instance of P_n is detected as

$$\left\{t_n^P \middle| \begin{array}{l} t_n=\max\{\omega_1(t_n)\},\\ t_n\in\text{Mid Stance.}\end{array}\right\}. \quad (7)$$

Initial condition of the joint angle $\theta_{i,n}^{ini}$ is calculated by only accelerometry in short calculation area δs. Considering a body segment movement in space, postural displacement is calculated as inclination to the gravity direction. The accelerometer is attached to the origin of body fixed frame $\{i\}$ that is fixed at the distal point of the segment (Fig. 5(a)). Acceleration measured by the accelerometer is simply expressed as

$${}^ia_{i,o}={}^i_0R\,{}^0a_{i,o}={}^i_0R(g+{}^0\ddot{Q}_i). \quad (8)$$

Where, ${}^ia_{i,o}$ represents an acceleration acting on the origin of body fixed frame $\{i\}$ (lower right suffix) with respect to frame $\{i\}$ (upper left suffix). ${}^i_0R$ is a rotational matrix that describes the orientation of the ground fixed frame $\{0\}$ with respect to the body-fixed frame $\{i\}$. ${}^0Q_i$ is a position vector of the origin of the frame $\{i\}$ with respect to the frame $\{0\}$. During the mid stance, ${}^0Q_i$ is assumed to be negligible and angle ψ_i is calculated as a relative inclination to the direction of gravity.

$$\psi_i=\mathrm{Atan2}(r_{21},r_{11}),\ {}^i_0R=\begin{bmatrix} r_{11} & r_{12}\\ r_{21} & r_{22}\end{bmatrix}. \quad (9)$$

Atan2 represents four-quadrant inverse tangent. An equivalent acceleration at position c_i is given by

$${}^ia_{i,c_i}={}^ia_{i,o}+{}^i\dot{\omega}_i\times{}^iP_{c_i}+{}^i\omega_i\times({}^i\omega_i\times{}^iP_{c_i}). \quad (10)$$

${}^i\omega_i$ is an angular velocity measured by the gyroscope. ${}^iP_{c_i}$ is the position vector of c_i. When body segments i and $i+1$ are linked with pin joint (Fig. 5(b)), their posture can be described by the calculation of the relative angle. Acceleration acting on the origin of frame $\{i+1\}$ can be described with respect to frame

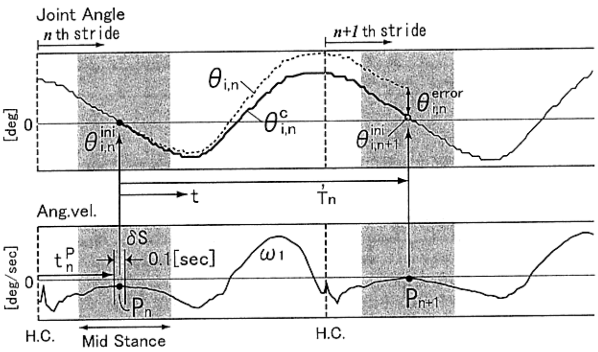
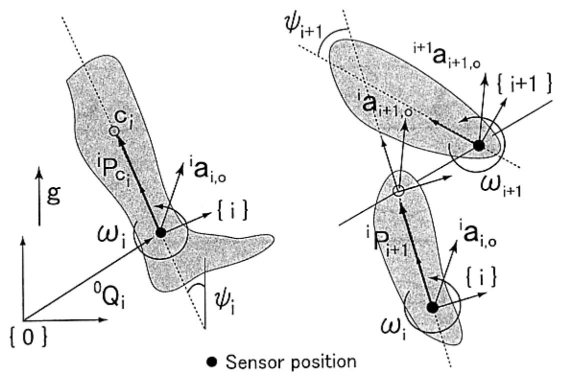


Fig. 4 Schematics of joint angle calculation. $\theta_{i,n}$ is joint angle calculated by Eq.(6) (broken line). $\theta_{i,n}^c$ is joint angle (solid line) compensated by Eq.(15). The point P_n is detected from ω_1 in mid stance (shaded) to determine integration time $\acute{T}_n$.



(a) (b)

Fig. 5 Rigid-body rinked models for the angle calculation by accelerometry. (a) A model for segment inclination calculation, (b) A model for relative angle calculation.

$\{i\}$ as follows.

$${}^{i}a_{i+1,o}={}^{i}a_{i,o}+{}^{i}\dot{\omega}_{i}\times{}^{i}P_{i+1}+{}^{i}\omega_{i}\times({}^{i}\omega_{i}\times{}^{i}P_{i+1}) \quad (11)$$

Where,

$${}^{i+1}a_{i+1,o}={}_{i}^{i+1}R\ {}^{i}a_{i+1,o} \quad (12)$$

We can identify ${}_{i}^{i+1}R$ by comparing the measured acceleration ${}^{i+1}a_{i+1,o}$ and the calculated equivalent acceleration ${}^{i}a_{i+1,o}$. Then, the joint angle is calculated by

$$\psi_{i+1}=\mathrm{Atan2}(r_{21},\ r_{11}),\ {}_{i}^{i+1}R=\begin{bmatrix} r_{11} & r_{12} \\ r_{21} & r_{22} \end{bmatrix} \quad (13)$$

Initial condition of the joint angle $\theta_{i,n}^{ini}$ is determined as

$$\theta_{i,n}^{ini}=\mathrm{mean}\ \{\psi_{i}(\delta s)\} \quad (14)$$

$$\delta s=[t_{n}^{P}-0.05,\ \cdots,\ t_{n}^{P}+0.05].$$

However, the calculation process given in Eq.(6) often causes integration drift error $\theta_{i,n}^{\mathrm{error}}$. To solve this problem, integration drifts are removed assuming that the error accumulate linearly to time progress t in integration time $\acute{T}_{n}$ (see, Fig. 4). The compensated angle $\theta_{i,n}^{c}$ can be calculated as follows.

$$\theta_{i,n}^{c}=\theta_{i,n}-(\theta_{i,n}^{\mathrm{error}}/\acute{T}_{n})t. \quad (15)$$

By using $\theta_{i,n}^{c}$, the accumulated drift errors can be successfully removed in every stride.

3.4 Inverse dynamic analysis

Iterative Newton-Euler dynamic algorithm is applied to solve the inverse dynamics problem[18],[19]. The algorithm is based on recursive formulation of force and moment balance equations of a free-body diagram. Newton-Euler dynamic algorithm is valid for open kinematic chains. Therefore, estimation of the total ground reaction force exerted on each of the leg is required. During mid stance in single support phase, the total ground reaction force is simply calculated by Eqs.(16) and (17). The acceleration of the left leg ${}^{0}a_{left,o}$ is not measured, but estimated assuming bilateral symmetry of the normal gait.

$$F_{rf}^{MST}=\sum_{i=1}^{5}m_{i}{}^{0}a_{i,ci}(t),\ t\in\text{Mid Stance} \quad (16)$$

$${}^{0}a_{left,o}(t)={}^{0}a_{right,o}(t+T/2). \quad (17)$$

Where, m_{i} is mass of the segment i, and ${}^{0}a_{i,ci}$ is the acceleration at the center of mass of the segment. T is a mean gait cycle calculated from the H.C. instance for all strides. Acceleration at the center of mass can be calculated using Eq.(10) with known body parameters. To obtain a smooth waveform, F_{rf}^{MST} is processed with a moving average filter. On the other hand, cubic spline function is applied to interpolate a sequence of desired intermediate data in the terminal stance and the double support phase. Cubic spline function defines continuous second derivatives of data. Joint moment τ_{i} is obtained by solving the equations of motion. Joint power P_{i} and energetic consumption E_{i} as a mechanical work are calculated as follows.

$$P_{i}=|\tau_{i}\,\omega_{i}|,\ E_{i}=\int P_{i}dt. \quad (18)$$

4. Result

The proposed temporal parameter detection algorithm is validated by comparing with the responses of force plates and the foot marker displacements. True H.C., T.O. instance and mid stance are determined from the rise of vertical ground reaction forces, and True H.O. instance is determined from vertical displacements of optical markers at heel. The detected event timings by our proposed method are compared with the true event timings (see, Fig. 6). Error is calculated as averaged difference of the detected instance (see, Table 1). The point A, B, C, and D have good agreement (Correlation coefficient : $r=0.98$) for H.C. SSPb, H.O. and T.O. respectively, although point A is always lagging behind the true H.C. at maximum of 0.12 sec. These results show that the proposed method can successfully detect precise temporal parameters according to the foot movement.

The calculated kinematic variables, joint angles, ground reaction forces and joint moments in a gait cycle are compared with those from the conventional

Table 1 Accuracy of temporal parameters determined by the proposed method in various cadences. Error is represented as mean value of absolute difference with standard deviation given in parenthesis.

Cadence [step/min]	True error. Mean (SD) [sec]			
	A - H.C.	B - SSPb	C - H.O.	D - T.O.
60	0.12(0.03)	-0.04(0.02)	0.04(0.09)	-0.19(0.12)
80	0.12(0.02)	-0.01(0.02)	0.05(0.02)	0.03(0.03)
100	0.11(0.01)	-0.03(0.01)	0.04(0.02)	0.01(0.01)
120	0.09(0.02)	-0.02(0.01)	0.02(0.03)	0.02(0.04)

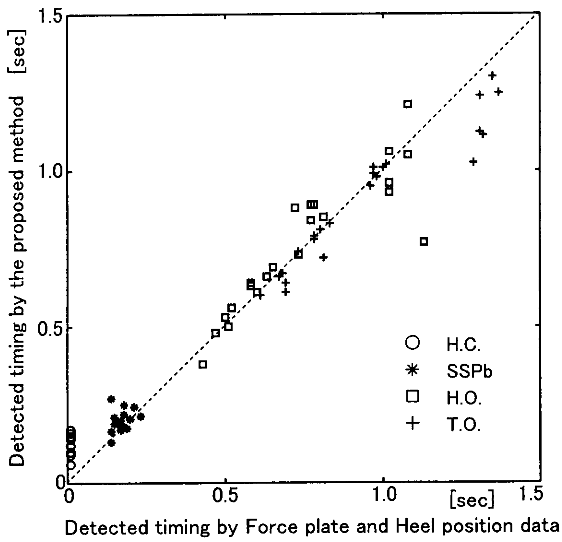


Fig. 6 Comparison of the detected foot contact event timings for a typical subject (24 steps including 4 cadences). The event timing is presented as time progress from true H.C. instance. Corelation coefficient is 0.98.

Table 2 Accuracy of joint angles, ground reaction forces, and joint moments calculated by the proposed method. Ground reaction force, and joint moment are normalized by body weight of the subject.

Cadence [step/min]	RMS of calculation error. Mean (SD) Joint angle [deg] Hip	Knee	Ground reaction force [N/BW] Horizontal	Vertical	Joint moment [Nm/BW] Hip	Knee
60	5.24(0.27)	11.22(1.09)	0.043(0.010)	0.23(0.054)	0.81(0.27)	0.57(0.08)
80	2.98(0.14)	8.96(1.10)	0.056(0.017)	0.27(0.028)	1.02(0.69)	0.55(0.25)
100	2.91(0.16)	8.83(1.14)	0.059(0.042)	0.28(0.044)	1.03(0.67)	0.54(0.22)
120	3.50(0.60)	9.73(0.69)	0.076(0.031)	0.31(0.012)	0.96(0.59)	0.58(0.23)

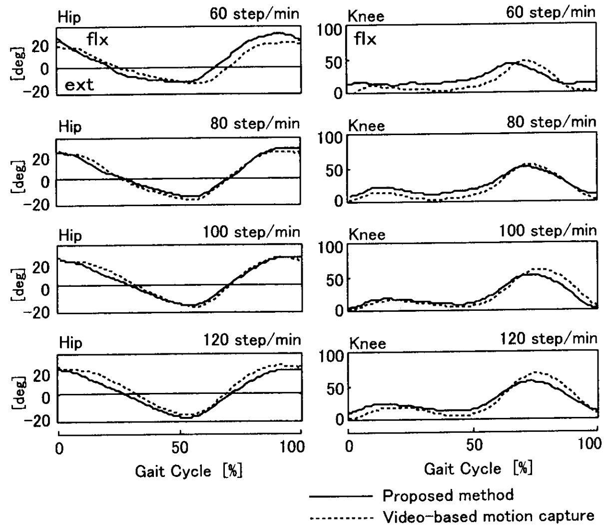


Fig. 7 A typical result of joint angles calculated the by proposed method for a gait cycle (solid line), comparing with the conventional standard methods (broken line).

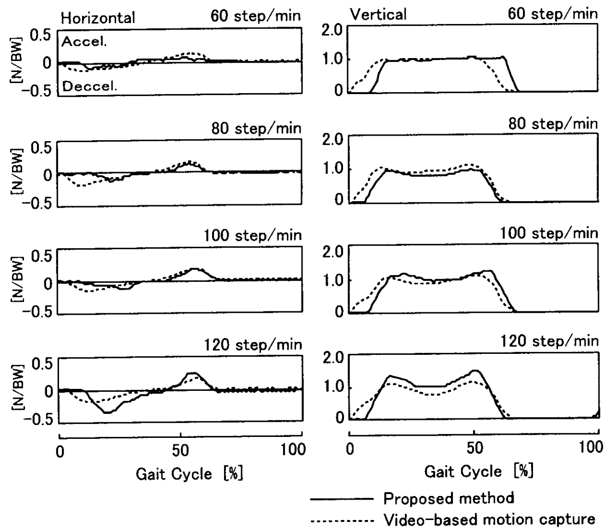


Fig. 8 A typical result of ground reaction force divided by body weight for normalization.

standard method. In both methods, five-segments rigid-body linked model is applied, assuming the ground reaction force is acting on the ankle joint. Error is expressed as average root mean square value (*RMS*) for all subjects (see, Table 2). *RMS* is calculated as follows

$$RMS=\sqrt{\sum_{n=1}^{N}\frac{(\hat{x}_n-x_n)^2}{N}}. \tag{19}$$

The notation n represents number of data points in a gait cycle. $\hat{x}_n$ is the true value from the conventional standard method, and x_n is the value calculated by the proposed method. These kinematic variables are almost consistent, but some deviations are clearly visible (Figs. 7 - 10). The accuracy of the joint angle calculation is decreased in both slowest walking (cadence 60 step/min) and fastest walking (cadence 120 step/min) with maximum error of 11.22 deg at knee joint. Error in the ground reaction force seems to be caused by inaccurate determination of the temporal parameters especially H.C.. Joint moments also agree well although largely effected by the error of ground reaction force via inverse dynamic calculation. The energy consumption during a gait cycle is also successfully evaluated (Fig. 11). Both tendencies

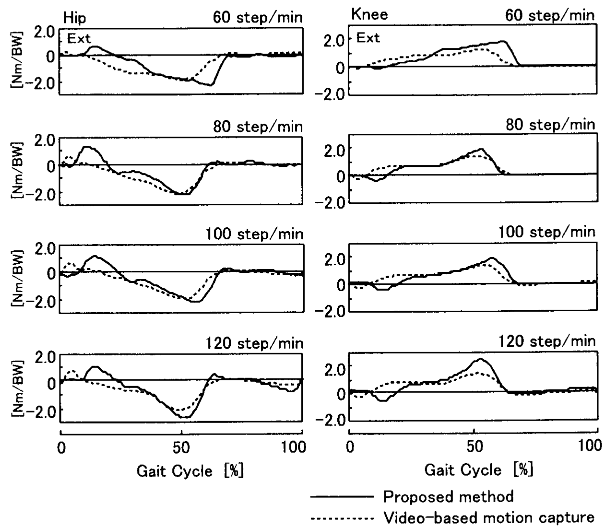


Fig. 9 A typical result of joint moments calculated by the five segments rigid-body linked model.

according to the increase of walking cadence agree well ($r=0.95$). These results show that the method can investigate fundamental locomotive dynamics and total mechanical energy consumptions reasonably.

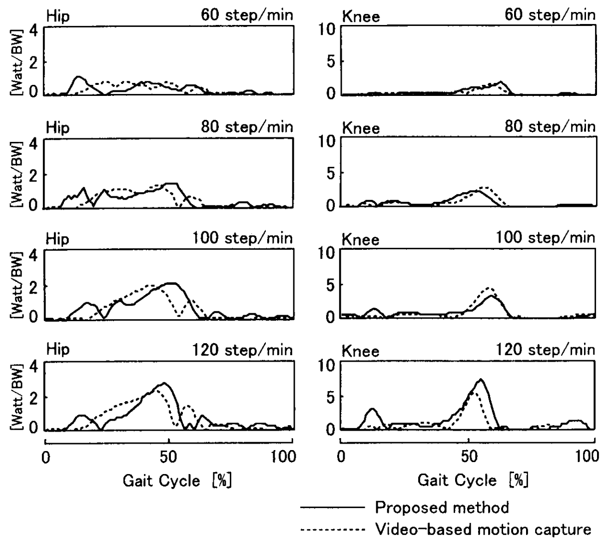


Fig. 10 A typical result of joint power divided by body weight for normalization.

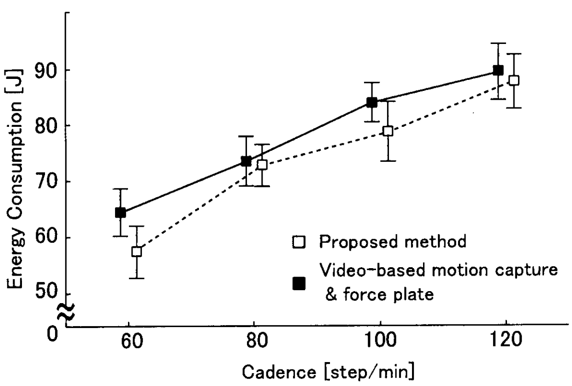


Fig. 11 A typical result of energy expenditure calculated as mechanical work for a gait cycle in variety of cadences, comparing with the result of the conventional method. Both tendencies agree well.

5. Discussion

A new method for the quantitative gait assessment is described. We have developed a small body-mounted instrument utilizing accelerometers and gyroscopes, which enable easy and smart kinematical calculation of locomotion. Numerous important gait parameters i.e. temporal parameters, joint angles, joint moments, and mechanical energy consumptions are calculated. The proposed method is verified by comparing the results with synchronized video-based motion capture system and force plates, evaluating error and sensitivity in variety of cadence of normal walking.

Errors in kinematic variables are mainly affected by the accuracy of temporal parameter determination and joint angle calculation. In temporal parameter determination, large errors are observed in slow walking. These errors are caused by the featureless acceleration waveforms due to slow motion. Moreover; the detection accuracy of H.C. is low regardless of walking speed. This suggests that the rise of distinct peak by ground contact and true H.C. differ from each other, and may needs some other appropriate procedure for precise H.C. detection. The minimum error is 0.01 sec. This suggests that the time resolution is also a major part of the error, which corresponds to the sampling period 0.01 sec. Accuracy is expected to improve by increasing the sampling rate and time resolution.

In angle calculations, there are two potential error sources. One is the error of the model assumptions in slow walking. In slow walking, increase of body movement in coronal plane is observed. This movement is unlikely to fulfill the assumption of the two-dimensional model in sagittal plane, and may cause decrease of accuracy. The other one is the effect of external fixation of the sensor unit in fast walking. It causes big abruption in acceleration measurement due to soft tissues, and results in inaccurate calculation of the initial condition of joint angle given in Eq.(14). This error can be dominated by improve the sensor attachment design for rigid sensor fixation. Special mathematical methods are also required to eliminate the soft tissue influence on the measurement(20). This problem needs further research.

The present study confirms the application of the proposed method to normal gait assessment. The method is design to be sufficiently functional even if only one side of the legs can be instrumented. However, the present algorithm requires necessitates some assumptions of normal gait such as bilateral symmetry. In order to raise the flexibility of the system, it is necessary to apply a more complicated calculation model without losing the convenience of measurement.

6. Conclusion

In this article, we have described an alternative method of quantitative gait analysis especially for the field studies with the completely body-mounted recording system. The method is based on the combination of direct measurement of dynamics of locomotion and a simple rigid-body linked model for kinematical calculation. Experiments have shown that the method can be used for continues measurement of the important gait parameters with easy and accurate kinematic calculations in normal level walking. However, the validation of the method has been limited to normal walking. We will extend this method to cover other types of human locomotion. Points to which special attention should be paid are that the proposed method does not require any special laboratory equipments, and can achieve total kinematical analysis provides useful information for the ability of locomotion. Further application of the proposed method may help to assess personal risk of falls in daily living environments.

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