Long-Term Unrestrained Measurement of Stride Length and Walking Velocity Utilizing a Piezoelectric Gyroscope

Shinji Miyazaki

***Abstract*—Long-term monitoring of stride length and walking velocity is considered to provide useful information for making decisions on treatment of patients with gait disabilities. The purpose of this study was to develop a device with the following design criteria: lightweight, easy attachment, little hindrance to the natural gait pattern, sufficient memory to record for one day, and practicality in clinical use. The prototype consists of a piezoelectric gyroscope, which detects angular velocity of the thigh of one leg in the sagittal plane, and a microprocessor-based maximum/minimum detector/data logger of a cyclic analog signal associated with the gait cycle. The accuracy of the device was evaluated in 20 normal subjects, seven above-the-knee (A/K) amputees, and ten hemiplegic patients, and relative accuracy within ±15% was obtained, except for two special cases.**

***Index Terms*—Ambulatory monitoring, gait, gyroscope, stride length, velocity.**

I. Introduction

REHABILITATION of patients with walking disabilities involves various professional staff and decision-making processes. Physiatrists evaluate the effectiveness of pharmacological and surgical treatments, physical therapists select appropriate therapeutic methods, prosthetists/orthotists prescribe and make necessary adjustments to appropriate devices, and medical social workers devise schedules for discharging patients at the earliest suitable time. In order to improve the quality of such decision making, attempts have been made to use technologies of gait analysis. The most commonly employed method involves the use of sophisticated systems, such as force plates and three-dimensional camera systems. However, measurements in a special gait laboratory with such space bounded devices tends to be artificial and of limited duration, thus, limiting the clinical value of the data obtained.

Unrestrained ambulatory monitoring of walking is more objective and enables measurements to be recorded in a natural setting during routine daily activities. Several ambulatory monitoring systems for recording foot pressure distribution have been developed [1]–[3], and monitors for recording joint angles [4] and load application to the pylons of artificial limbs [5], [6] have been reported. The problems with these devices include limited memory capacity and consequently, short recording times [1]–[4], a high total weight [2], [3], and limited fields of application [5], [6]. The purpose of this study was to develop a general ambulatory monitor for clinical usage with the following criteria: light weight, easy attachment, little hindrance to the natural gait pattern, sufficient memory to record for one day, and clinically useful presentation of the data obtained. The stride length and walking velocity were selected as measurement parameters because they provide the most basic functional information about the gait. Stride length in this paper refers to the sum of the left and right step lengths. Several researchers have used these parameters to study pathological gait [7]–[11]. In this paper, the general system configuration is described first, followed by an explanation of the algorithm used for stride/velocity measurement, and the results of accuracy evaluations in normal and patient populations are presented.

II. System Configuration

Fig. 1 shows the configuration of the system, which consists of an angle transducer, a general purpose min/max monitor/logger [12], and a personal computer.

The angle transducer consists of a piezoelectric gyroscope and an integrator. The gyroscope is attached to the thigh and detects its angular velocity in the sagittal plane. The integrator converts the angular velocity to motion angle. The monitor is based on an Intel Z80 compatible CMOS one-chip microprocessor (Z8T4C015-10; Toshiba Manufacturing Co. Ltd., Tokyo, Japan). It detects the maximum and minimum values of an analog signal associated with the gait cycle, which is the motion angle in the present case, and logs them and the cycle time for each gait cycle. Fig. 2 shows how the monitor works. First, cyclic gait signal is compared to a reference voltage v_{ref} and converted to a digital pulse. A level shift circuit is incorporated in the transducer part so that the gait signal changes cyclically above and below V_{ref}. The microprocessor detects the rising edges of the digital pulse, and their interval, i.e., cycle time, is converted to 2-b digital number using clock pulses which run at 5 kHz. The maximum peak during the cycle is detected by a peak holder circuit, and the deviation of the peak from the reference voltage $\Delta V_{\max}$ is converted to a digital pulse of which the width is proportional

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The author is with Institute for Medical and Dental Engineering, Tokyo Medical and Dental University, 2-3-10, Kanda-Surugadai, Chiyoda-ku, 101, Tokyo, Japan (e-mail: miyazaki@elec.i-mde.tmd.ac.jp).

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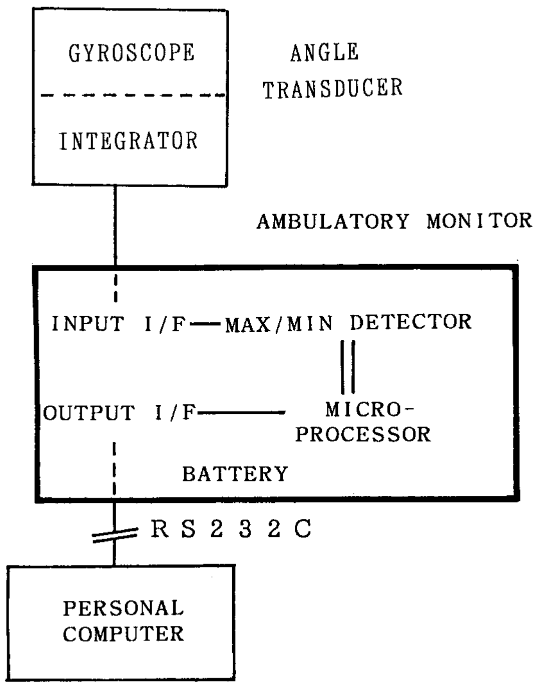


Fig. 1. General configuration of the system.

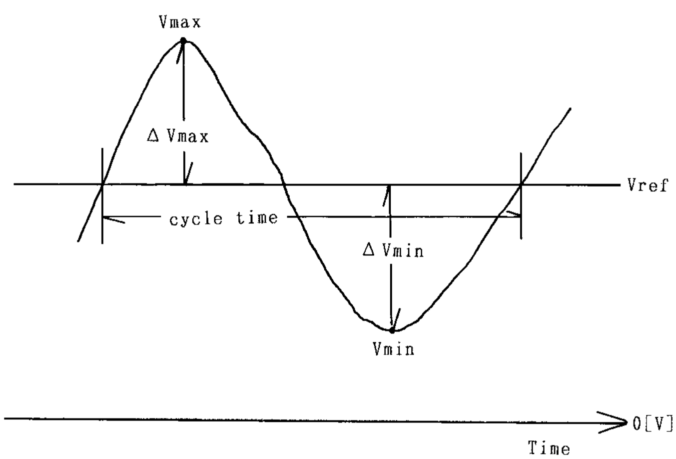


Fig. 2. Schema of min/max detection of general gait monitor.

to $\Delta V_{\max}$. The pulse width is converted to one-byte digital number by the microprocessor using other clock pulses which run at 300 kHz. Similarly, the deviation of the minimum peak from the reference voltage $\Delta V_{\min}$ is converted to one-byte digital number. Thus, four bytes are used to describe one gait cycle. The maximum number of cycles which can be stored by the monitor is 8000, which is sufficient to log gait data for one day. The monitor is operated by two 9-V alkaline batteries, measures 140 (W) $\times$ 80 (H) $\times$ 40 (D) mm and its total mass, including batteries and a plastic case, is 150 g (Fig. 3).

When the necessary data have been stored by the monitor, they are transferred to a personal computer via an RS232C cable at 9600 baud. Any personal computer with an RS232C interface can be used, and it takes approximately 40 s to transfer the data of 8000 cycles.

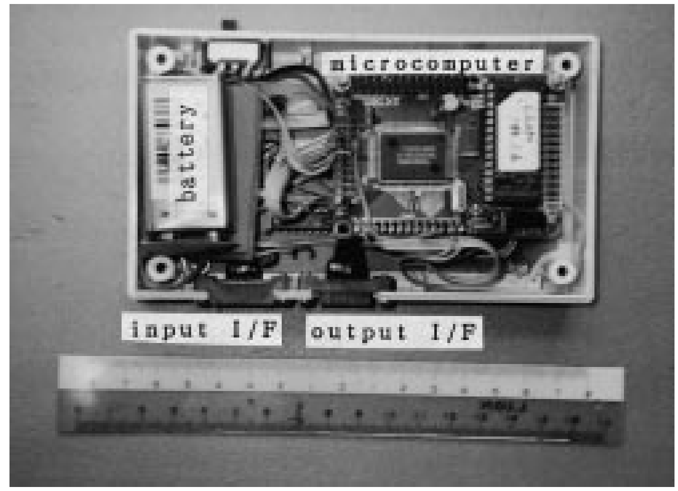


Fig. 3. Photograph of the monitor unit.

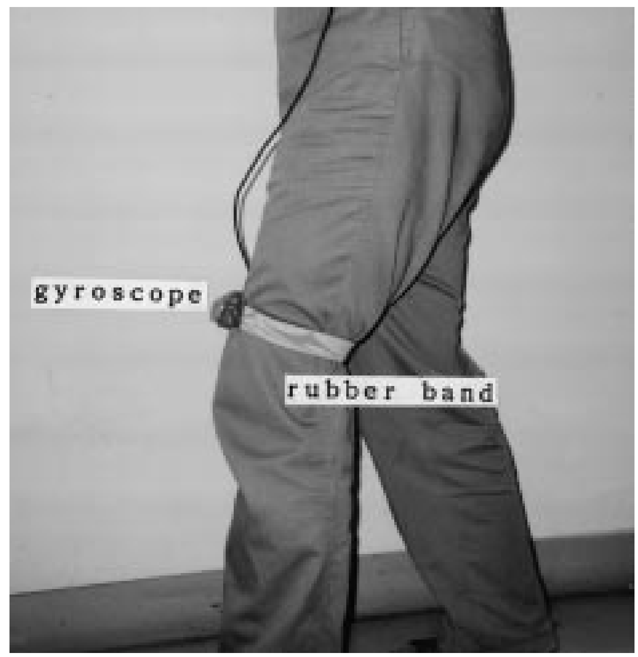


Fig. 4. Attachment of the gyroscope sensor to the thigh.

III. Method of Measuring Stride Length and Walking Velocity

A. Gyroscope

A small light weight gyroscope based on the piezoelectric principle (ENV-05S; Murata Manufacturing Co. Ltd., Kyoto, Japan), measuring $24 \times 24 \times 58$ mm and weighing 45 g was used. This gyroscope detects angular velocity about its own longitudinal axis, and its measurement range is $\pm 150°$/s. It is attached to the frontal aspect of the lower part of the thigh with a rubber band with its longitudinal axis aligned perpendicular to the sagittal plane, as shown in Fig. 4.

B. Gait Model

In order to measure the stride length, the angle of the thigh with respect to the absolute space is required. The

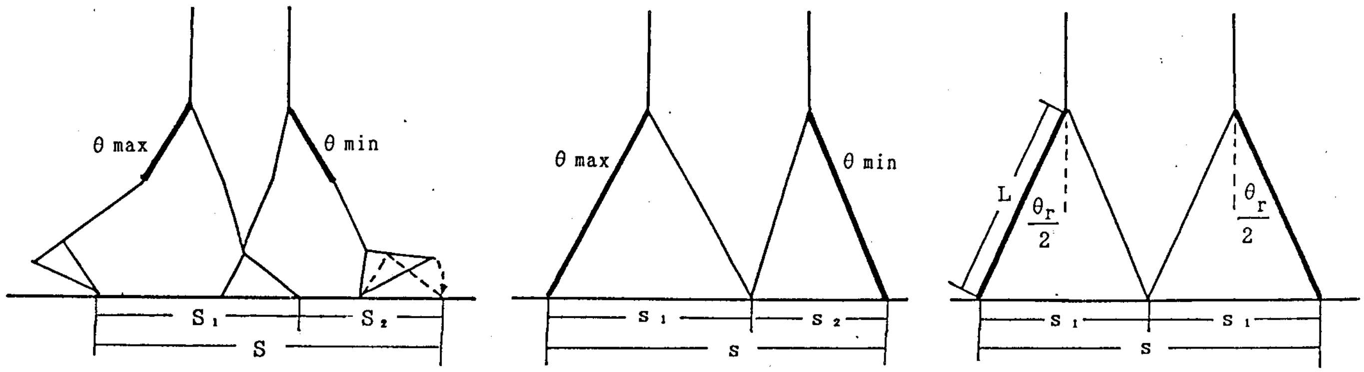


Fig. 5. Gait model for calculating stride length.

gyro scope output, however, contains dc drifts, and complete integration, either by hardware or software, will cause errors in the angle or even saturation, so an operational-amplifier-based integrator with slight leakage is used to avoid this. The effective time constant of the leakage decay is 2.0 s. Thus, only the relative amplitude of the absolute angle, i.e., θ_r, defined as the difference between the maximum $\theta_{\max}$ and minimum $\theta_{\min}$, is meaningful (θ_r can be obtained from the max/min detecting function of the general monitor by summing $\Delta\theta_{\max}$ and $\Delta\theta_{\min}$).

Therefore, although the absolute angles of the thigh, shank, and foot of both legs are required to calculate the stride length S precisely, as shown in Fig. 5(a), a very simple symmetric gait model of extended limbs, as shown in Fig. 5(c), was employed to calculate the approximate value of the stride length s. Using this simplified gait model, the approximate step length of the first step s_1 is given by

$$s_1 = 2 \times L \times \sin\left(\frac{\theta_r}{2}\right) \tag{1}$$

where L is an extended leg length. Since a symmetry gait is assumed, the step length of the second step s_2 is equal to s_1. Thus, the stride length s is given by

$$\begin{aligned} s &= s_1 + s_2 \\ &= 2 \times s_1. \end{aligned} \tag{2}$$

C. Instantaneous Walking Velocity

The precise cycle time C for each gait cycle can be extracted by the max/min detector which operates with a quartz clock. The instantaneous walking velocity V is defined as

$$V = \frac{S}{C}. \tag{3}$$

IV. Evaluation of Measurement Accuracy

Dependence of approximate stride length value on walking velocity.

In order to find out how close the approximate stride length s given by (2) is to the true stride length S, the following experiments were carried out. It is obvious that the accuracy of the instantaneous velocity corresponds to that of the stride length. A normal healthy subject, referred to as subject A below (age 46 yr, height 170 cm), was asked to walk on a straight, level 40 m walkway at various velocities as naturally as possible with the gyroscope attached to his right thigh. The total time C^t to cover this distance was calculated by summing cycle times C of all cycles. The mean walking velocity V^* was calculated as

$$V^* = \frac{40}{C^t}. \tag{4}$$

The mean true stride length S^* was calculated as

$$S^* = \frac{40}{N} \tag{5}$$

where N is the total number of strides. A preliminary experiment in which each true stride length S as obtained from the foot prints was compared to each approximate stride length s revealed that the measurement error for each s value, i.e., the difference between S and s, did not differ significantly from the difference between S^* and s^*, where s^* is defined as the mean of s calculated using (2) for all the strides. Therefore, in the following, it is assumed that the accuracy of each stride can be estimated by the error in s^*.

Fig. 6 shows the ratio of s^* to S^* as a function of the mean walking velocity V^*. This ratio will be one if s^* is equal to S^*, but its values were actually greater than one because s^* was greater than S^*, and the difference between them was greater at lower mean walking velocities. The broken line in Fig. 6 is a linear regression line calculated for all the data. This ratio was also calculated for common asymmetric gait disabilities. A hemiplegic gait was adopted for this purpose, and normal subject A simulated the three point reciprocal gait of patients with left-sided hemiplegia. The gyroscope was again attached to his right thigh, i.e., the sound side. The results are shown in Fig. 6 by open circles, and the dotted line is the linear regression line.

A. Compensation and Calibration of Stride Length and Velocity

As shown in Fig. 6, the approximate stride length s depends roughly linearly on the mean walking velocity V. Therefore, if this dependency on V is known, s can be compensated for

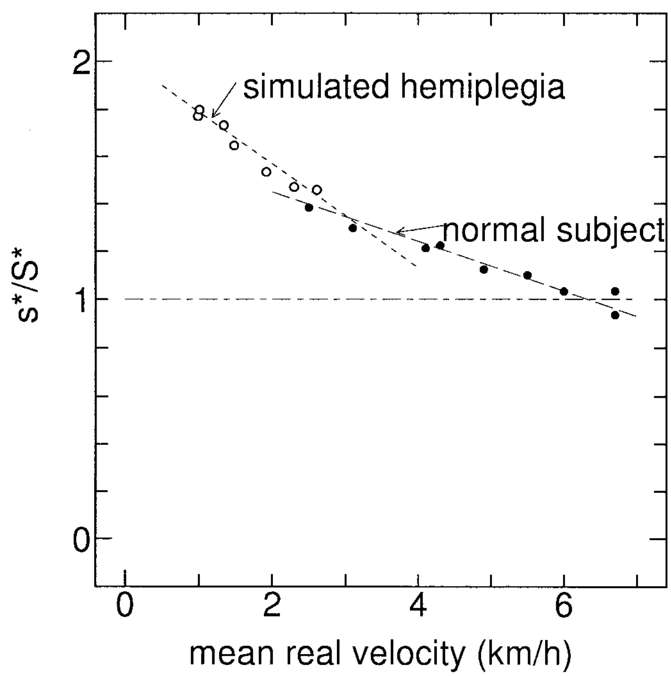


Fig. 6. Dependence of stride length on walking velocity. The vertical axis represents the ratio of s^* (mean approximate stride) to S^* (mean true stride). Filled and open circles represent data for normal and the simulated hemiplegic gait, respectively. Broken and dotted lines shows linear regression lines for the normal and hemiplegic data, respectively.

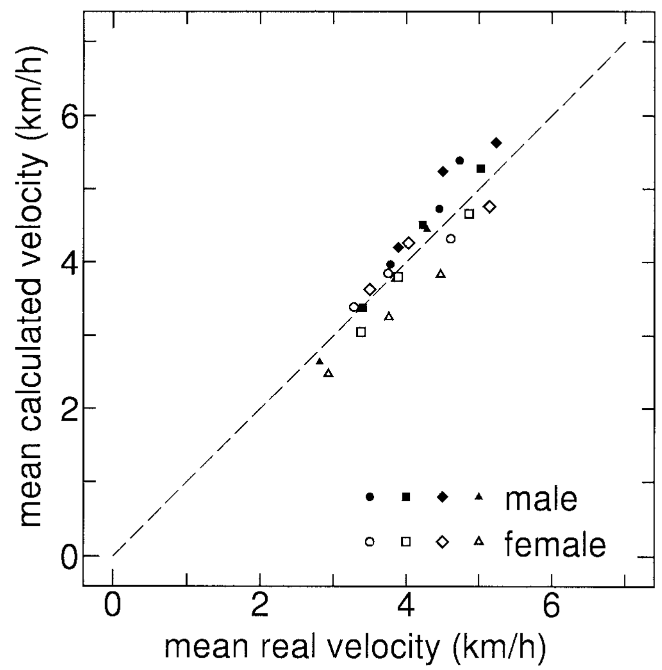


Fig. 7. Compensation of velocity dependence. Filled and open symbols represent male and female subjects, respectively.

it, and a closer estimate can be calculated. The dependence is assumed to be described by

$$\frac{s}{S} = aV + b \tag{6}$$

and the relationship between V and S is given by

$$V = \frac{S}{C}. \tag{7}$$

Similarly, the relationship between the approximate stride length s and the true cycle time C is given by

$$v = \frac{s}{C} \tag{8}$$

where v is the approximate instantaneous walking velocity. From (6)–(8), the following equation can be derived:

$$v = aV^2 + bV. \tag{9}$$

Therefore, if the approximate stride length is calculated using (2) and the value of v is calculated using (8), a closer estimate of V can be obtained by solving the second-order equation of V and reassigned as v', which represents the compensated walking velocity value. Then compensated value stride length s' can be obtained by substituting v' into the left side of (7)

$$s' = Cv'. \tag{10}$$

In order to assess the power of such compensation, seven normal male (age range 20–83 yr; mean 37) and 11 female (age range 19–77 yr; mean 42) subjects were asked to walk 25 m on a straight level floor as naturally as possible, with the gyroscope attached to the right thigh. Measurements were recorded three times at a moderate walking velocity, twice at a high velocity, and finally, twice at a low velocity for each subject. Four male and four female subjects were selected arbitrarily, and one datum for each walking velocity was also arbitrarily selected. Using the linear regression data for normal walking of subject A, the data for each of these subjects were compensated for velocity dependence. For the actual compensation, the regression line was recalculated with respect to the real velocity normalized by the limb length. The results are shown in Fig. 7. The horizontal axis represents the mean real velocity V^* defined by

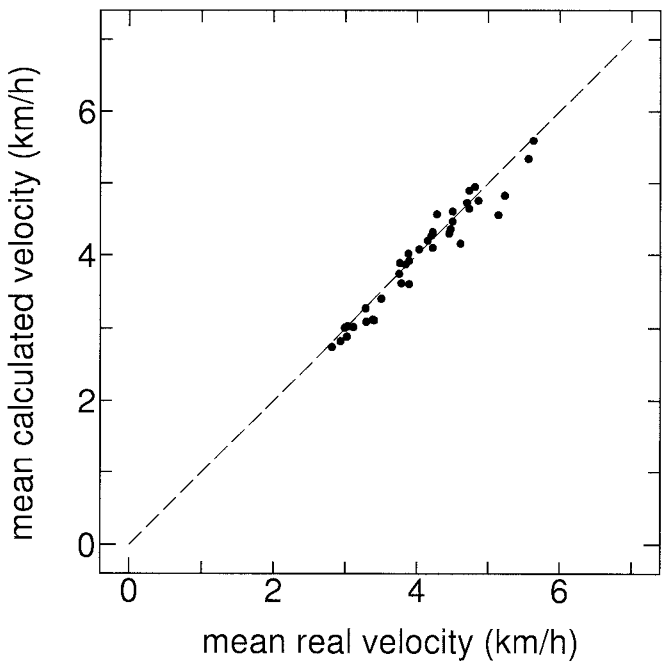
$$V^* = \frac{25}{C^t}. \tag{11}$$

The vertical axis represents the mean calculated velocity v^* defined by

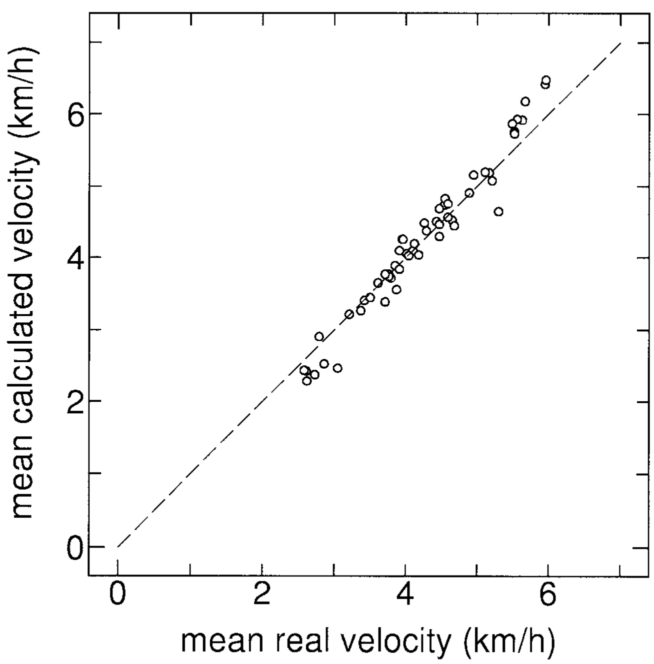
$$v^* = \frac{\Sigma v'}{N} \tag{12}$$

where N is the total number of strides and the summation is done for all strides. The calculated velocity compares fairly well with the real velocity, with the data for the male and female subjects tending to lie above and below the 45° broken line representing the ideal case. This tendency may be attributable to the different gait patterns of the male and female subjects.

Close examination of the data shown in Fig. 7 reveals that there is a general tendency for small values to be obtained at the other two velocities when a small value is obtained after compensation at one velocity. So, after compensating for the velocity dependence using the linear regression line of subject



(a)



(b)

Fig. 8. Effects of subject-dependent calibration: (a) eight male subjects and (b) 11 female subjects.

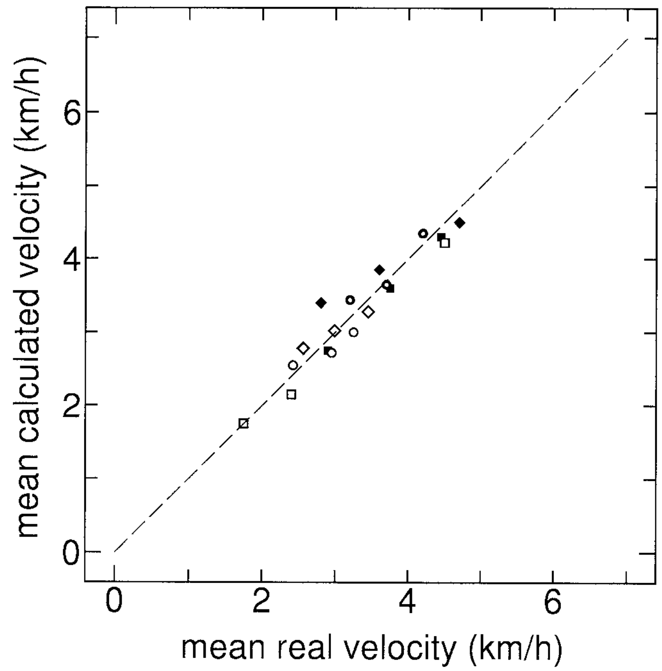


Fig. 9. Results of seven A/K amputees.

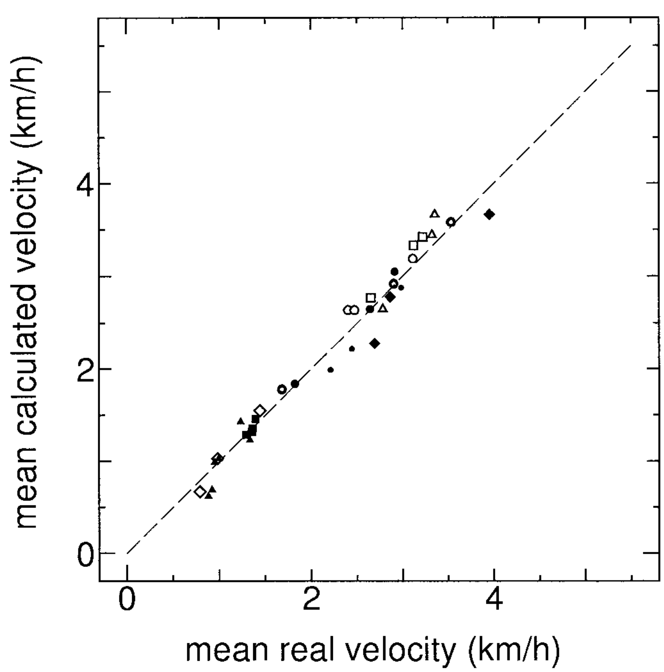


Fig. 10. Results of ten hemiplegic patients.

A, one datum at the moderate walking velocity was selected and a scaling factor k was calculated so that

$$v^* = kV^* \tag{13}$$

at that specific velocity. Other data for other trials at moderate, high, and low velocities are then re-scaled using the scaling factor, thus obtained. The effects of this subject-dependent calibration is summarized in Fig. 8. Data for all male and female subjects are shown in Fig. 8(a) and (b), respectively. The measurement accuracy was improved further by this subject-dependent calibration.

B. Results with Patients' Data

In order to assess the effectiveness of the procedures described above, data were collected from seven patients with A/K prostheses and ten hemiplegic subjects. The experimental methods and sequence were same as those described for the normal subjects, except 1) the number of walking trials

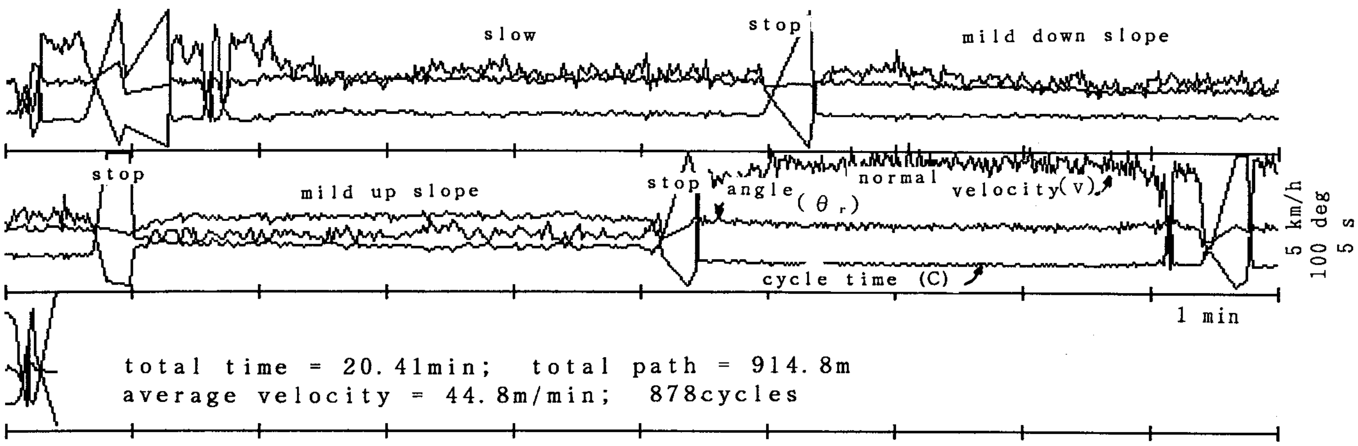


Fig. 11. Example of computer display of the secondarily processed data for outdoor walking by an A/K amputee. Data taken in sequential cycles are connected by straight lines for the convenience of easy comprehension by viewers. Thus datum after standstill creates ramp pattern as an artifact; for example, the cycle time of standstill cycle is extremely longer than the previous one.

was two at medium velocity and one each at high and low velocities, 2) the hemiplegic patients walked only 15 m, and 3) the gyroscope was attached to the thigh of the sound side. The compensation coefficients for dependence on walking velocity, a) and b), were calculated from the normal gait of subject A for the amputees. For the hemiplegic subjects, they were calculated from the normal gait of subject A when their medium walking velocity was higher than 2 km/h, and from the simulated hemiplegic gait when it was below than 2 km/h. Fig. 9 and 10 show the results of the amputees, and the hemiplegic subjects, respectively. Mean real velocity and mean calculated velocity are defined in a similar manner as in the case of healthy subjects except that 25 in (11) is replaced by 15 in the case of hemiplegic subjects. Even though the patients' gaits were asymmetrical, fairly good estimation of the walking velocities were achieved. The maximum relative error was within ±15% for both categories of gait disabilities, except for two hemiplegic subjects, for whom the maximum relative errors were about −25%. These two patients walked in a three-point following gait, i.e., they walked with a cane and took steps with the sound limb first and then moved the affected limb to that point later. This may be the reason why their maximum relative errors were relatively large, as the simulated hemiplegic gait of subject A was a three-point reciprocal gait.

V. Example of Secondary Processing

The above results were all obtained on a straight level floor in a building. As an example of secondary processing of data obtained outdoors, Fig. 11 shows the hard copy of the color display of a personal computer's monitor. An A/K amputee walked on the pavement of a street for about 20 min. He was asked to stand still for about 30 s when he came to places where the walking conditions changed, e.g., slow to normal, or a level surface became a mild slope. On the original display, the cycle time, relative thigh angle (θ_r), and instantaneous walking velocity are shown by yellow, blue, and red lines, respectively, for each walking cycle.

VI. Discussion

An ambulatory system for monitoring the stride length and instantaneous walking velocity was developed. Conventional methods of measuring the stride length using a switch matrix [10], [11] or center of pressure (c.o.p). obtained from force plates impose physical restraints on the placement of the monitor. To the author's knowledge, no system which imposes no restraints on subjects during long-term gait measurements has been developed so far. The cycle time can also be measured by the present system. This feature will be of clinical importance, since the major reason for an increased final walking velocity may be an increased stride length and/or decreased walking cycle duration. Thus, discerning the reason would provide additional information for making clinical decisions, e.g., fitting of a prosthesis or orthosis.

One may think that more information, e.g., asymmetry of gait, would be obtained if two gyroscopes, one on each limb, are used. However, there are at least two good reasons why only one gyroscope is used; 1) due to dc drift inherent in the gyroscope, there is no way to calculate the true absolute angle of the limb with respect to the gravity and thus, calculation of the true stride length is impossible and 2) by the same token, there is no good reason to add further physical restrains on the patient by using two gyroscopes and thus, two cables on both limbs, since there is no way to measure true step length of left and right limbs separately.

The present system utilizes a relatively simple, light, small gyroscope as the sensor and offers fairly good measurement accuracy. The primary reason for this is that when processing the raw data, knowledge based on the velocity dependence of the calculated stride length and a personal calibration technique are employed. The latter calibration can be carried out without imposing a load on the subject as it can be done while the subject is walking a known distance on a walkway, which can be a general purpose indoor floor or a path outdoors.

When the system is utilized as an unrestrained gait monitor, an additional device to record the conditions under which the subject walks may be necessary. A commercially available microcassette audio tape recorder could be used for this purpose, and the subject could record his/her announcements about the conditions. Even with this additional information, however, there is no suitable means of calibrating gait patterns which differs from the original ones made on a flat level floor, e.g., walking up and down slopes and steps. Only in cases when the walking conditions are known in advance, can several separate calibrations be performed to obtain reasonably accurate results. Bearing these problems in mind, it is considered that the present system will prove a valuable tool for providing "additional" information that will be helpful for the various decision making processes for patients with gait disabilities. Further data processing, such as tracing and comparing the gait performances during the rehabilitation process, may be necessary to ensure that this system is really useful clinically.

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Shinji Miyazaki received the B.E., M.Sc., and Ph.D. degrees from the University of Tokyo, Tokyo, Japan, in 1971, 1973, and 1979, respectively. He was an exchanging graduate student at Drexel University, Philadelphia, PA, in the 1979 to 1980 academic year.

He has been working with the Institute for Medical and Dental Engineering, Tokyo Medical and Dental University, Chiyoda-ku, Tokyo, Japan, since 1973. He is currently a Research Instructor. His research interest lies in the field of rehabilitation engineering, with main focus on gait and motion analysis. He is also running new projects in the field of whole-body hyperthermia and neural prostheses.