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Spatio-temporal parameters of gait measured by an ambulatory system using miniature gyroscopes

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Abstract

In this study we describe an ambulatory system for estimation of spatio-temporal parameters during long periods of walking. This original method based on wavelet analysis is proposed to compute the values of temporal gait parameters from the angular velocity of lower limbs. Based on a mechanical model, the medio-lateral rotation of the lower limbs during stance and swing, the stride length and velocity are estimated by integration of the angular velocity. Measurement's accuracy was assessed using as a criterion standard the information provided by foot pressure sensors. To assess the accuracy of the method on a broad range of performance for each gait parameter, we gathered data from young and elderly subjects. No significant error was observed for toe-off detection, while a slight systematic delay (10 ms on average) existed between heelstrike obtained from gyroscopes and footswitch. There was no significant difference between actual spatial parameters (stride length and velocity) and their estimated values. Errors for velocity and stride length estimations were 0.06 m/s and 0.07 m, respectively. This system is light, portable, inexpensive and does not provoke any discomfort to subjects. It can be carried for long periods of time, thus providing new longitudinal information such as stride-to-stride variability of gait. Several clinical applications can be proposed such as outcome evaluation after total knee or hip replacement, external prosthesis adjustment for amputees, monitoring of rehabilitation progress, gait analysis in neurological diseases, and fall risk estimation in elderly. © 2002 Published by Elsevier Science Ltd.

Keywords: Gait analysis; Gyroscope; Wavelet analysis; Ambulatory system; Stride velocity

1. Introduction

Walking is one of the most common human physical activities. Evaluation of time and distance parameters during walking is helpful in assessing abnormal gait, to quantify improvement resulting from interventions, or to predict subsequent events such as falls. This evaluation is useful in several clinical situations such as: functional performance assessment after treatment or surgery such as hip and knee arthroplasty (Wall et al., 1981), refining proper alignment and fit of external prosthesis or orthesis (Czerniecki and Gitter, 1996), fall risk assessment in elderly persons (Maki, 1997), or selecting the appropriate assistive device.

A gait laboratory based on camera, walkway with implanted sensors (Hirokawa and Matsumara, 1987) or force-plates, and electromyography allows a complete gait analysis but require a dedicated laboratory. Although these techniques have been widely used for research purposes, their sophistication, time required for setting up the instrumentation and analyze the data, as well as cost, have hindered their use in clinical practice. In addition, these techniques require the subjects to walk in a pre-defined specific path, and assume that data measured from only a few steps are representative of usual gait performance. To avoid these limitations, ambulatory recording systems carried by the subject and allowing outdoor measurement have been developed, using new technology, such as a powerful microcontroller, miniature sensors, high capacity memory and small batteries.

Ambulatory systems using foot switches or pressure sensors attached to the sole are used to monitor

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temporal parameters (Zhu et al., 1991; Abu-Faraj et al., 1997). These techniques provide satisfactory results for normal walking, but sensor attachment proves difficult when assessing subjects with abnormal gait. Even when correctly done, several problems, such as shuffling gait, mechanical failure, or patient acceptance limit their applicability. Another system using insole with a matrix of multitude pressure sensors has a high reliability to evaluate the time of each phase of the gait cycle. Although this system provides additional information on foot pressure distribution during gait, it cannot be used for a long period of measurement, and sensors do not provide information during the swing phase.

Recently, technical progress has made possible the realization of miniature kinematic sensors such as accelerometers and angular rate sensors with integrated conditioning and calibrating module (Sparks et al., 1998). In addition, due to their very low consumption, these sensors can be battery powered and are promising tools for long-term ambulatory monitoring. The possibility of detecting simple parameters such as steps and cycle time from trunk or heel acceleration has already been demonstrated using miniature accelerometers (Evans et al., 1991; Aminian et al., 1995). More recently the efficiency of accelerometers to detect temporal features of walking has been shown (Aminian et al., 1999). However, the acceleration signal is often biased by the gravitational acceleration and depends on the attachment site along the segment (e.g. limb). Since the actual nature of walking consists of lower limb rotation around joint articulation, the use of miniature angular rate sensors (gyroscopes) has proven to be an alternative technique for gait analysis (Tong and Granat, 1999; Nene et al., 1999). However, no clear correspondence has been established between gait events and the patterns of angular velocity. This is probably because gait events (e.g. heelstrike and toe-off) are transitory signals that cannot be properly enhanced by traditional signal processing. We decided to use wavelet transform to overcome this problem because this technique has proven its superiority over traditional filtering techniques in simple and complex kinematic data (Ismail and Asfour, 1999).

In this study, we describe an ambulatory system for estimation of spatio-temporal parameters during long periods of walking. An original method, based on wavelet transform is proposed to compute the values of gait parameters from the angular velocity of lower limbs. The accuracy of measurements was assessed, using as a criterion standard the information provided by foot pressure sensors. In order to estimate the accuracy of the method on a broad range of performance for each gait parameter, data were gathered from young and elderly subjects.

2. Method

2.1. Gyroscopes

Lower limbs movement during walking was measured using miniature gyroscopes. The gyroscope consists of a vibrating element coupled to a sensing element, acting as a Coriolis sensor. Coriolis effect is an apparent force that arises in a rotating reference frame and is proportional to the angular rate of rotation. The piezoelectric gyroscope measures the Coriolis acceleration generated when a rotational angular velocity is applied to the oscillating piezoelectric bimorph. Three miniature piezoelectric gyroscopes (Murata, ENC-03J) were used. Because of their low current consumption, these sensors can be battery powered and are therefore appropriate for ambulatory monitoring. The signal from each gyroscope was amplified and low-pass filtered to remove the electronic noise of the gyroscope. Each gyroscope and the conditioning electronic were placed in a small circular box ($D = 25$ mm, $H = 10$ mm) (Fig. 1). A swiveling electromechanical system was designed to calibrate and characterize each sensor. Various angular velocities were applied to the gyroscope. An optical sensor measured the time corresponding to perform a rotation of 360° and allowed calibration of the gyroscope. The amplifier gain of each gyroscope was set to obtain a sensitivity of 4 mV/deg/s. The gyroscopes were attached with a rubber band to each shank and on the right thigh. Each sensor measured the angular rate rotation parallel to the mediolateral axis (perpendicular to the sagittal plane). The signals were digitized (12 bit) at a sampling rate of 200 Hz by a portable data logger (Physilog, BioAGM, CH) and stored on a memory card. At the end of the recording the data were transferred to the computer for analysis.

2.2. Footswitches

Footswitches have been used as criterion standard to detect the gait events and validate the gyroscope measuring method. The footswitches consisted of two pairs of force sensing resistors, FSR (Interlink, LU)

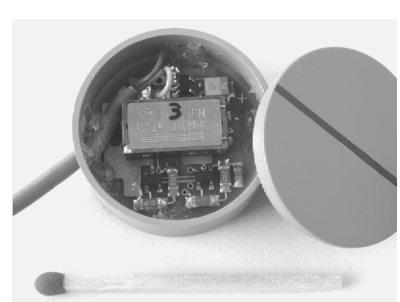

Fig. 1. Gyroscope module with embedded electronic. The gyroscopes are attached on shank segments and right thigh.

Fig. 2. Footswitches (FSR) attachment beneath the heel (two sensors) and beneath the big toe (two sensors).

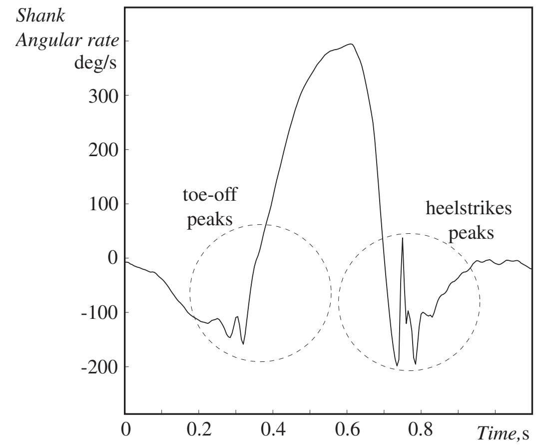

Fig. 3. Shank angular velocity showing the presence of peaks during the toe-off and heelstrike.

placed under the right foot (Fig. 2): FSR_{heel} beneath the heel (two sensors) and FSR_{toe} beneath the big toe (two sensors). For each pair, the sensors were connected in series and the outputs were simultaneously digitized and recorded by *Physilog* with those from the gyroscopes at a sampling rate of 200 Hz. During heelstrike and toe-off, FSR_{heel} and FSR_{toe} signals are subjected to abrupt changes. By thresholding the derivative of these two signals the exact time of heelstrike and toe-off was obtained.

2.3. Subjects

Measurement using gyroscopes and FSR were taken from a group of 9 young (21 ± 2 yr) and 11 elderly (79 ± 8 yr) subjects. Each young subject performed four trials including at least 20 gait cycles. The first three trials were performed on a treadmill at the preferred speed, under and over the preferred speed, respectively. The fourth trial was performed with a comfortable velocity over ground on a 30 m long walkway. The elderly performed only two trials on the walkway at their comfortable velocity. The study protocol was accepted by the institutional review board (Faculty of Medicine, University of Lausanne). Written informed consent was obtained from each subject.

2.4. Wavelet analysis: gait temporal parameters estimation

In order to compute the temporal parameters such as the duration of swing, single and double stances during a gait cycle, it is necessary and sufficient to determine for each leg the precise moments of heelstrike (when the foot first touches the floor) and toe-off (when it takes off) during that cycle. These events have distinctive signal features of shank angular velocity appearing as rather sharp negative peaks involving some medium and relative high frequencies (Fig. 3). Although the amplitude of these peaks vary according to various parameters such as subject's velocity or weight, or the presence of limping due to a painful articulation, they can always be localized as long as one knows where to look for it in time and frequency domains. To successfully achieve this task, we have decomposed the shank angular velocity into wavelet packages. Wavelet transformation is well adapted for gait events identification because it allows detection of a specified frequency at a specified time (e.g. toe-off and heelstrike). In other words, toe-off and heelstrikes events consist of combined features that can be well located in the time and frequency range. The methods of analysis need to be versatile enough to handle gait events that can be at opposite extremes in terms of their time–frequency location. Moreover, wavelets transformation allows the use of a suitable wavelet function (Coiflet wavelet), which is more similar to the pattern of heelstrike and toe-off.

A multi-resolution wavelet decomposition (Mallat, 1989) was used to enhance the toe-off and heel-strike event during walking. This technique consists in splitting a signal into high-scale (low-frequency components) called *approximation* and low-scale (high-frequency components) called *detail*. The decomposition is iterated, with successive approximations (or details) being decomposed in turn, so the original signal is broken down into many lower-resolution components. At each decomposition scale the number of samples in the time domain is decreased by throwing away every second sample (down sampling with a factor of '2'). By considering the original signal $s(n)$ (shank angular velocity), the approximation of the signal at scale $j = 0$

is $A_{2^0}s$ which corresponds to the original signal $s(n)$. At each new decomposition the approximation and detail decrease in resolution since at each scale 2 samples will be replaced by one. At scale j the $A_{2^j}s$ represents approximation of $s(n)$ with a resolution of one sample for every 2^j samples of the original signal. We have

$$s = A_{2^1}s + D_{2^1}s = A_{2^2}s + D_{2^2}s + D_{2^1}s = \cdots = A_{2^j}s + D_{2^j}s + D_{2^{j-1}}s \cdots + D_{2^1}s. \quad (1)$$

Thus

$$A_{2^j}s = A_{2^{j+1}}s + D_{2^{j+1}}s. \quad (2)$$

By using a suitable low-pass filter h, and a high-pass filter g, the approximate signal $A_{2^{j+1}}s$ and detail signal $D_{2^{j+1}}s$ can be further written as following:

$$A_{2^{j+1}}s = \sum_{k=-\infty}^{+\infty} h(2n-k)\, A_{2^j}s, \quad (3)$$

$$D_{2^{j+1}}s = \sum_{k=-\infty}^{+\infty} g(2n-k) A_{2^j}s. \quad (4)$$

Furthermore the signal can be reconstructed from approximate and detail signal using the following transform:

$$A_{2^j}s = 2\sum_{k=-\infty}^{+\infty} \tilde{h}(n-2k) A_{2^{j+1}}s + 2\sum_{k=-\infty}^{+\infty} \tilde{g}(n-2k) D_{2^{j+1}}s, \quad (5)$$

where

$$\tilde{h}(n) = h(-n), \quad (6)$$

$$\tilde{g}(n) = g(-n). \quad (7)$$

The coefficients of h and g filter are associated with the shape of wavelet considered for the analysis. In this study, a decomposition into 10 scales with the Coiflet wavelet of order 5 has been used.

The flowchart for the estimation of toe-off and heelstrikes are shown in Fig. 4. For clarity, let us consider only one shank. It is obvious that the algorithm will be applied in the same way for both shanks. First an approximation of $s(n)$ corresponding to the $s_a = \sum_{j=1}^{9} D_{2^j}s$ was obtained. With this approximation the original signal was only considered in the range of 0.14–36 Hz. Drift and high frequency movement artifacts were canceled in this way. Then two new approximations were obtained. The first one enhanced the heelstrike component ($A_{2^1}s_a - A_{2^9}s_a$) and the second one enhanced the toe-off component ($A_{2^3}s_a - A_{2^9}s_a$). For each of these approximations, the time corresponding to the global maximum values of the signal (ms) were detected. These values corresponded roughly to the time of midswing during a gait cycle (see Fig. 5), however their exact significance is not of interest to us.

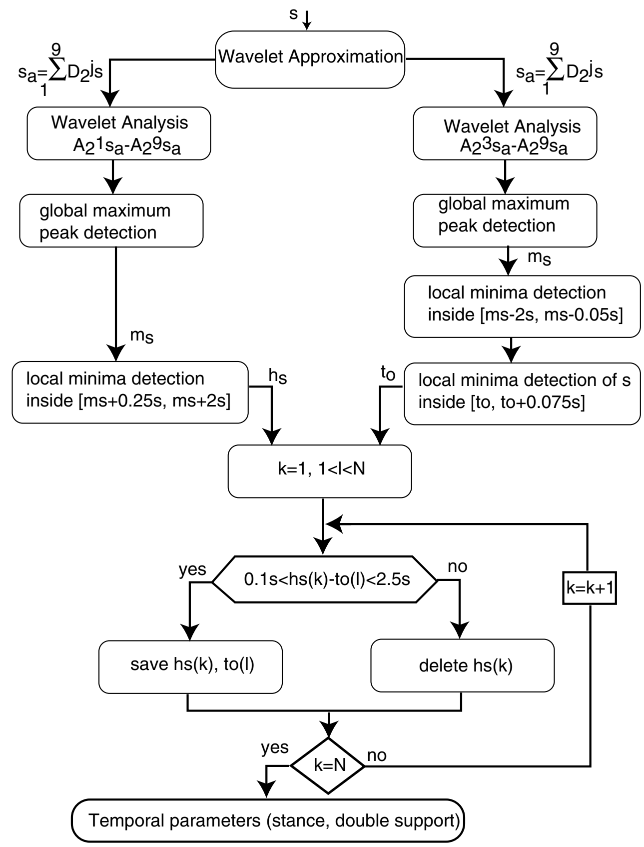

Fig. 4. Algorithm flowchart for temporal parameters estimation from shank angular rate signal (s).

They will only be used as reference to select the intervals in which negative peaks reminiscent of heelstrikes and toe-offs are to be found. So we next looked for local minima inside intervals [ms + 0.25s, ms + 2s] and [ms−2s, ms−0.05s], assuming that a gait cycle is always less than 2 s. The result was two different series of samples hs (heelstrikes) and to (toe-offs). The toe-off detection was refined by finding the local minima inside of the original signal s in the intervals [to, to + 0.075s]. Finally, at each cycle, the actual heelstrike and toe-off must be so that their difference is considered reasonable for walking. Thus, for each hs(k) only the nearest to(l) verifying the condition $0.1s < \text{hs}(k) - \text{to}(l) < 2.5s$ was saved. If no to(l) met this condition, $h(k)$ was deleted and the next heelstrike was considered. Once we are in possession of hs_L (left heelstrike), to_L (left toe-off), hs_R (right heelstrike) and to_R (right toe-off), every temporal parameter of a gait cycle can be easily computed as percentage of gait cycle. These parameters are:

- Duration of each *gait cycle* (measured at right or left leg)

$$\text{RGCT}(k) = \text{to}_R(k+1) - \text{to}_R(k) \quad \text{or}$$
$$\text{LGCT}(k) = \text{to}_L(k+1) - \text{to}_L(k), \ 1 \leqslant k \leqslant N. \quad (8)$$

N: *number of gait cycles*

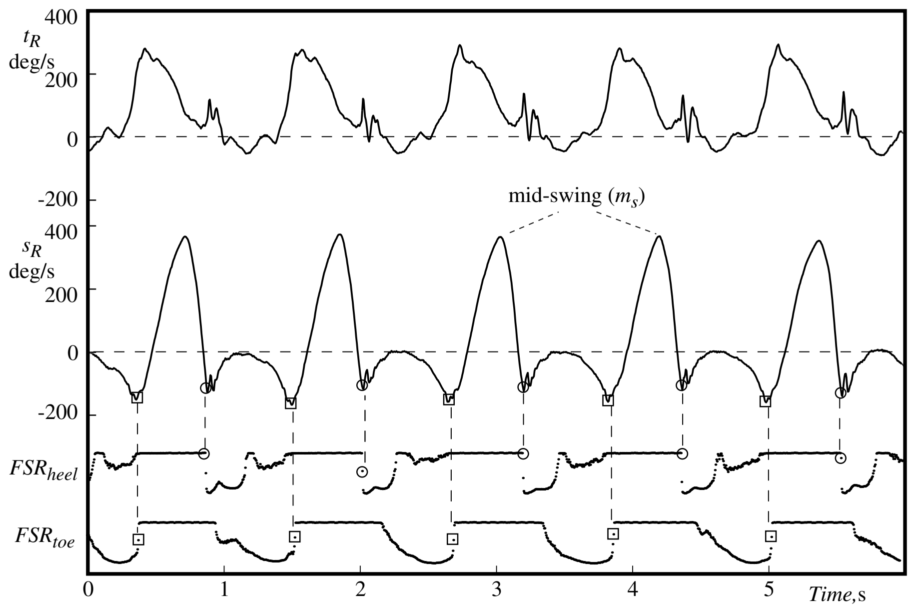

Fig. 5. FSR signal from heel (FSR_{heel}) and toe (FSR_{toe}), angular rate signal at right shank (s_R) and right thigh (t_R). Heelstrike (○) and toe-off (□) detected by FSR and s_R are shown on each signal during 5 typical gait cycles. m_s correspond approximately to the time of mid-swing.

- *Left stance* (time between left heelstrike and left toe-off)

$$\mathrm{LS}(k) = \mathrm{to_L}(k) - \mathrm{hs_L}(k). \quad (9)$$

- *Right stance* (time between right heelstrike and right toe-off)

$$\mathrm{RS}(k) = \mathrm{to_R}(k) - \mathrm{hs_R}(k). \quad (10)$$

- *Initial double support* (time between right heelstrike and left toe-off), known also as left double thrust support time (Wall et al., 1981)

$$\mathrm{IDS}(k) = \mathrm{to_L}(k) - \mathrm{hs_R}(k). \quad (11)$$

- *Terminal double support* (time between left heelstrike and right toe-off), known also as right double thrust support time (Wall et al., 1981)

$$\mathrm{TDS}(k) = \mathrm{to_R}(k+1) - \mathrm{hs_L}(k). \quad (12)$$

- Double support

$$\mathrm{DS}(k) = \mathrm{IDS}(k) + \mathrm{TDS}(k). \quad (13)$$

In order to reach a good precision for the temporal parameters, at least 20 gait cycles have been considered for analysis ($N > 20$). For each trial, the value of temporal parameters was averaged over the N cycles. Statistical analysis is then performed to find the significance and accuracy of the parameters obtained by gyroscopes in comparison with the FSR-based system. The root mean square error (RMSE) was used

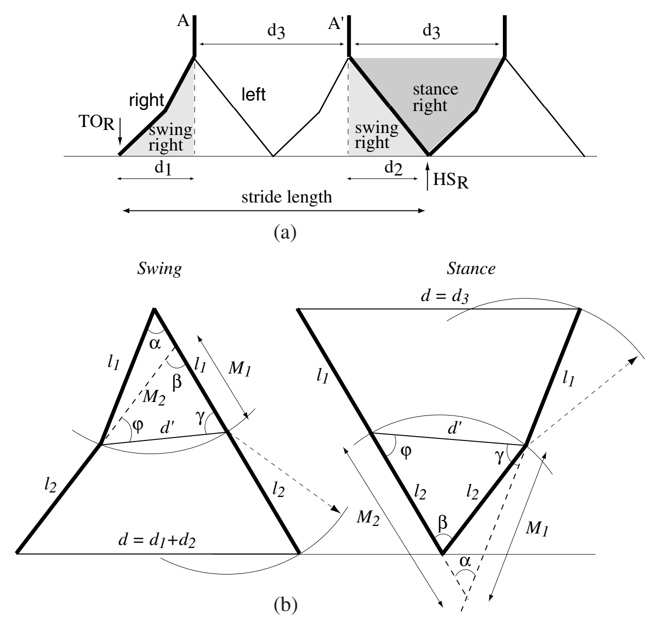

Fig. 6. (a) Body segments model during a gait cycle started by right toe-off ($\mathrm{to_R}$) and ended by right heelstrike ($\mathrm{hs_R}$). (b) Shank and thigh configuration during stance and swing phase. Thigh and shank length (l_1 and l_2, respectively) are measured on each subject. α and β obtained from angular rate signals (t_R and s_R) allow to estimate the distance d during stance and swing phases.

to analyze the average difference between estimated data by gyroscope and actual data obtained by FSR.

$$\mathrm{RMSE} = \sqrt{\frac{\sum_{\text{number of samples}} (\text{estimated} - \text{actual})^2}{\text{number of samples}}}. \quad (14)$$

2.5. Gait model: spatial parameters estimation

We propose a double segment gait model involving both shank and thigh. In this model, the swing phase is considered as a double pendulum model, while the stance phase is considered as an inverse double pendulum model. Consequently, knowledge of stance and swing phases obtained from the method described in Section 2.4 is necessary. Fig. 6(a) illustrates the rotation of the shanks and thighs during a complete gait cycle. The distance $d_1 + d_2$ corresponds to the part of stride length performing by the swing of right shank and right thigh. In addition, during this right swing phase, the body has moved forward by the distance d_3 involved by the rotation of left shank and left thigh throughout the contralateral stance phase. Assuming the same length for left and right steps, one can consider that the rotation during left stance phase is equal to rotation during right stance phase. Stride length (SL) can then be measured as follows:

$$\mathrm{SL} = d_1 + d_2 + d_3. \tag{15}$$

By considering α and β to be the angular rotation of the right thigh and right shank, respectively, one can calculate the distance $d = d_1 + d_2$ from the trigonometric relations (Fig. 6b).

During the swing phase we have for each gait cycle k:

$$d(k) = d_1(k) + d_2(k) = \sqrt{(l_2 + M_1(k))^2 + (l_2 + M_2(k))^2 - (l_2 + M_1(k))(l_2 + M_2(k))\cos\beta(k)}, \tag{16}$$

where, l_1 is the thigh length, l_2 the shank length

$$M_1(k) = \frac{\sin\varphi(k)}{\sin\beta(k)} d'(k), \tag{17}$$

$$M_2(k) = \frac{\sin\gamma(k)}{\sin\beta(k)} d'(k) \tag{18}$$

with :

$$\gamma(k) = \frac{\pi - \alpha(k)}{2}$$

and

$$\varphi(k) = \frac{\pi - 2\beta(k) + \alpha(k)}{2}$$

and

$$d'(k) = l_1\sqrt{2(1 - \cos\alpha(k))}. \tag{19}$$

The angles α and β were estimated by integration of the angular rate rotations of the right thigh (t_R) and right shank (s_R) as follows:

$$\alpha(k) = \int_{\mathrm{to_R}(k)}^{\mathrm{hs_R}(k)} t_\mathrm{R}(t)\,\mathrm{d}t,$$

$$\beta(k) = \int_{\mathrm{to_R}(k)}^{\mathrm{hs_R}(k)} s_\mathrm{R}(t)\,\mathrm{d}t. \tag{20}$$

During the stance phase we have

$$d(k) = d_3(k) = \sqrt{(l_1 + M_2(k))^2 + (l_1 + M_1(k))^2 - (l_1 + M_2(k))(l_1 + M_1(k))\cos\alpha(k)}, \tag{21}$$

where:

$$M_1(k) = \frac{\sin\varphi(k)}{\sin\beta(k)} d'(k), \tag{22}$$

$$M_2(k) = \frac{\sin\gamma(k)}{\sin\beta(k)} d'(k) \tag{23}$$

$$\begin{gathered} \text{with :} \\ \gamma(k) = \frac{\pi - \beta(k)}{2} \\ \text{and} \\ \varphi(k) = \frac{\pi - 2\alpha(k) + \beta(k)}{2} \\ \text{and} \\ d'(k) = l_2(k)\sqrt{2(1 - \cos\beta(k))} \end{gathered} \tag{24}$$

and

$$\begin{aligned} \alpha(k) &= \textstyle\int_{\mathrm{hs_R}(k)}^{\mathrm{to_R}(k)} t_R(t)\,\mathrm{d}t, \\ \beta(k) &= \textstyle\int_{\mathrm{hs_R}(k)}^{\mathrm{to_R}(k)} s_R(t)\,\mathrm{d}t. \end{aligned} \tag{25}$$

For each gait cycle k, the stride length (SL) and stride velocity (SV) were obtained as follows:

$$\mathrm{SL}(k) = d_1(k) + d_2(k) + d_3(k), \tag{26}$$

$$\mathrm{SV}(k) = \frac{\mathrm{SL}(k)}{\mathrm{RGCT}(k)}, \tag{27}$$

where RGCT(k) was obtained from (8). In order to evaluate the accuracy of stride length and velocity given by (26) and (27), the following measurements were carried out. For treadmill walking, the value of treadmill speed has been considered as the actual mean walking velocity (V_a). A preliminary experiment in which various treadmill speeds were compared to the speed of an odometer placed on the treadmill belt confirmed that the treadmill speed is accurate. During ground (or "non-treadmill") walking, the mean velocity was evaluated by considering the time needed to cover 20 m using a stopwatch. The actual mean stride length (L_a) was calculated as

$$L_a = V_a \, GCT_a, \tag{28}$$

where GCT_a is the mean gait cycle time obtained from footswitches. The accuracy of stride length and velocity was evaluated by comparing L_a and V_a with the mean of SL and SV (SL_m and SV_m) over 20 m.

3. Results

3.1. Temporal parameters

The correspondence between temporal events detected by FSR and the gyroscope pattern is shown in Fig. 5 for five consecutive gait cycles. Using the algorithm of Fig. 4, left and right heelstrikes (respectively toe-off) were detected from shank gyroscopes (Fig. 7). Then based on relation (8) to (13), the mean value (over the N cycles) of left and right gait cycle time ($LGCT_m$, $RGCT_m$), left and right stance (LS_m and RS_m), initial and terminal double stance (IDS_m, TDS_m, DS_m) were computed for each trial on treadmill and ground. Simultaneously, the periods of the actual gait cycle (GCT_a) and right stance (RS_a) were detected from FSR_{heel} and FSR_{toe}.

Fig. 8 compares stance period (RS_m and RS_a) and gait cycle times values ($RGCT_m$ and GCT_a) obtained from the gyroscopes to those of FSR from a trial performed by the young subjects (treadmill and ground and elderly during overground walking. A good agreement ($r > 0.99$) was found. The difference between RS_m and RS_a was not significant ($N = 58$, $p > 0.80$) with a mean standard error RMSE = 23 ms. No significant change was observed between $RGCT_m$ and GCT_a ($N = 58$, $p > 0.97$). The RMSE between $RGCT_m$ and GCT_a was 8 ms (0.5%). The 95% confidence interval for the difference between heelstrike detection by FSR and gyroscope was [7 ms, 13 ms]. Heelstrike detection based on the gyroscopes occurred later (10 ms in average) than FSR (underestimation). The 95% confidence interval for the difference between toe-off detection by FSR and the gyroscopes was [−5 ms, 4 ms]. Therefore there was no significant difference in toe-off detection. The RMSE between $LGCT_m$ and $RGCT_m$ was 15 ms ($r > 0.99$, 1.3%), which shows the validity of the method in providing the same time for both left and right gait cycles.

The mean values of right and left stance (RS_m and LS_m) as well as double supports (TDS_m, IDS_m, DS_m) estimated by the gyroscopes are shown in Table 1. In general, higher values are observed for elderly subjects.

3.2. Spatial parameters

For each trial the mean value of stride length and velocity (SL_m and SV_m) were estimated by finding the average of stride length (SL) and stride velocity SV (relation 26 and 27) over the 20 m of walking. Fig. 9 compares actual L_a and V_a to SL_m and SV_m. No significant change was found between L_a and SL_m ($N = 56$, $p > 0.52$) and neither between V_a and SV_m ($p > 0.66$). The coefficient of correlation is high in both cases ($r > 0.96$). Actual velocity and stride length can be estimated from the measured values as follows:

$$V_a = 1.02 SV_m - 0.04, \tag{29}$$

$$L_a = 0.97 SL_m. \tag{30}$$

The RMSE for velocity estimation is 0.06 m/s (6.7%) and that of stride length estimation is 0.07 m (7.2%).

Stride length and velocity had lower averaged values for the elderly compared to young subjects.

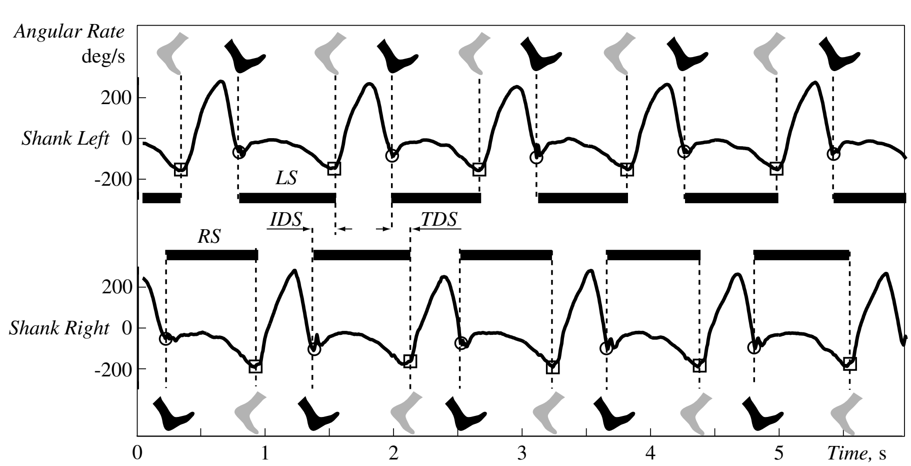

Fig. 7. Temporal parameters estimation from heelstrikes (○) and toe-offs (□) of each leg based on relations (8) to (13) in the text.

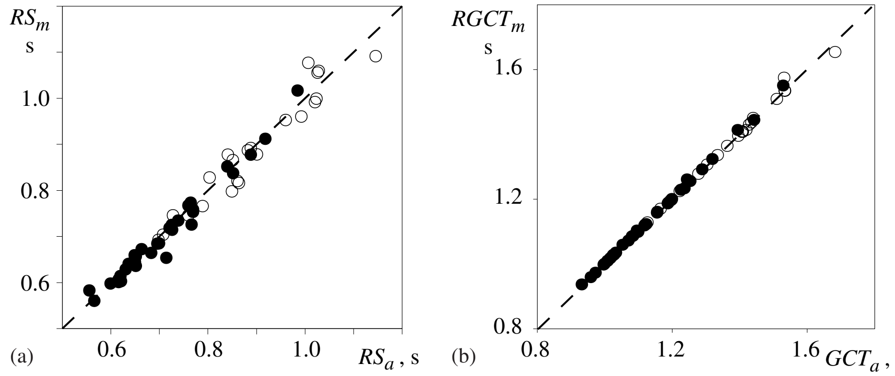

Fig. 8. (a) Right stance duration, in s, obtained from gyroscope (RS_m) compared with that obtained from pressure sensors (RS_a), $r > 0.99$. (b) Gait cycle time from gyroscope ($RGCT_m$) compared to actual gait cycle time obtained from FRS (GCT_a), $r > 0.99$. (●) young, (○) elderly subjects.

Table 1
The averaged values of right stance (RS_m), left stance (LS_m), initial double stance (IDS_m), terminal double stance (TDS_m) and double stance (DS_m) in the percentage of gait cycle obtained for young and elderly subjects

Subjects	RS_m (%)	LS_m (%)	IDS_m (%)	TDS_m (%)	DS_m (%)
Young	61 ± 2	62 ± 2	12 ± 2	12 ± 2	23 ± 3
Elderly	64 ± 3	65 ± 2	14 ± 3	15 ± 2	29 ± 4

Fig. 10 shows typical results obtained during 2 min of treadmill walking where the velocity was fixed at 1.11 m/s (4 km/h). The subject was asked to walk at his normal pace, to decrease his pace then increase his pace. All parameters were obtained at each gait cycle from gyroscopes. As it is illustrated, the estimated velocity (SV) is almost constant ($SV_m = 1.14$ m/s) with a coefficient of variation of 4%. The change of gait cycle (RGCT) provides a corresponding change of stride length (SL) in order to maintain the velocity constant. The corresponding coefficient of variation of SL and RGCT is 15%. During the same period there is no significant variation of left and right stance (coefficient of variation less than 3%) while the double support time (DS) has a coefficient of variation of 16%.

4. Discussion

We have proposed a new algorithm based on wavelet transformation to detect toe-off and heelstrike from shank angular velocity. A high correlation with actual gait events detected by foot pressure sensors was found. In addition, no significant error was observed for toe-off detection, while a slight systematic delay (10 ms in average) existed between heelstrike detected by gyroscopes compared to FSR. Considering a sampling period of 5 ms (200 Hz), this error is acceptable. The obtained values correspond to the established values for young and elderly populations which is around 60% for the duration of stance and 20% for the duration of double support (Rose and Gamble, 1994; Maki, 1997). Few studies have been reported about temporal gait parameters' detection based on gyroscope. Pappas et al. (1999) used a gyroscope on the heel in conjunction with FSR to detect the precise time of heel-off. Tong and Granat (1999) used a gyroscope on the shank and thigh. They found that the pattern on the shank has two minima, one occurs at foot flat and the other occurs at toe-off. This last result is consistent with ours. In contrast to their results, we found that the first peak after wavelet enhancement corresponds to heelstrike and not foot flat. It must be noted that only 2 subjects were involved in their study and the sampling rate was 50 Hz. Toe-off and heelstrike detection were based on the choice of a threshold on the FSR signal. It is clear that the value of the threshold can modify the results. Moreover, only a few steps were considered for analysis which cannot be considered as statistically significant.

The method presented in this study compares favorably with other methods used to identify gait events that require more complex instrumentation and a laboratory setting. Hreljac and Marshall (2000) used four motion analysis cameras (60 Hz), involving 12 trials of two healthy subjects and found an absolute average error of 4.7 and 5.6 ms for heelstrike and toe-off, respectively, considering information from a forceplate as the "gold standard". Stanhope et al. (1990) using a 50 Hz camera collection system, reported an error greater than 20 ms in over 20% of the cases. Using accelerometers on the thigh and considering FSR as reference, Aminian et al. (1999) found the 95%

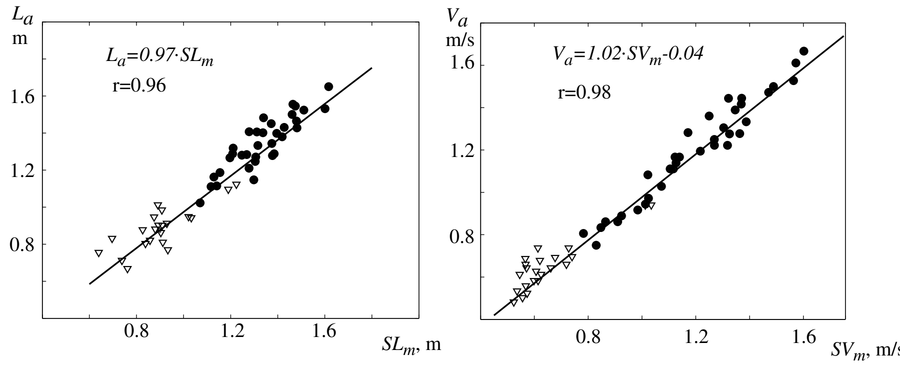

Fig. 9. Correspondence between actual stride length (L_a) and velocity (V_a) and estimated values from shank and thigh gyroscopes (SL_m, SV_m) for young (●) and elderly subjects (▽).

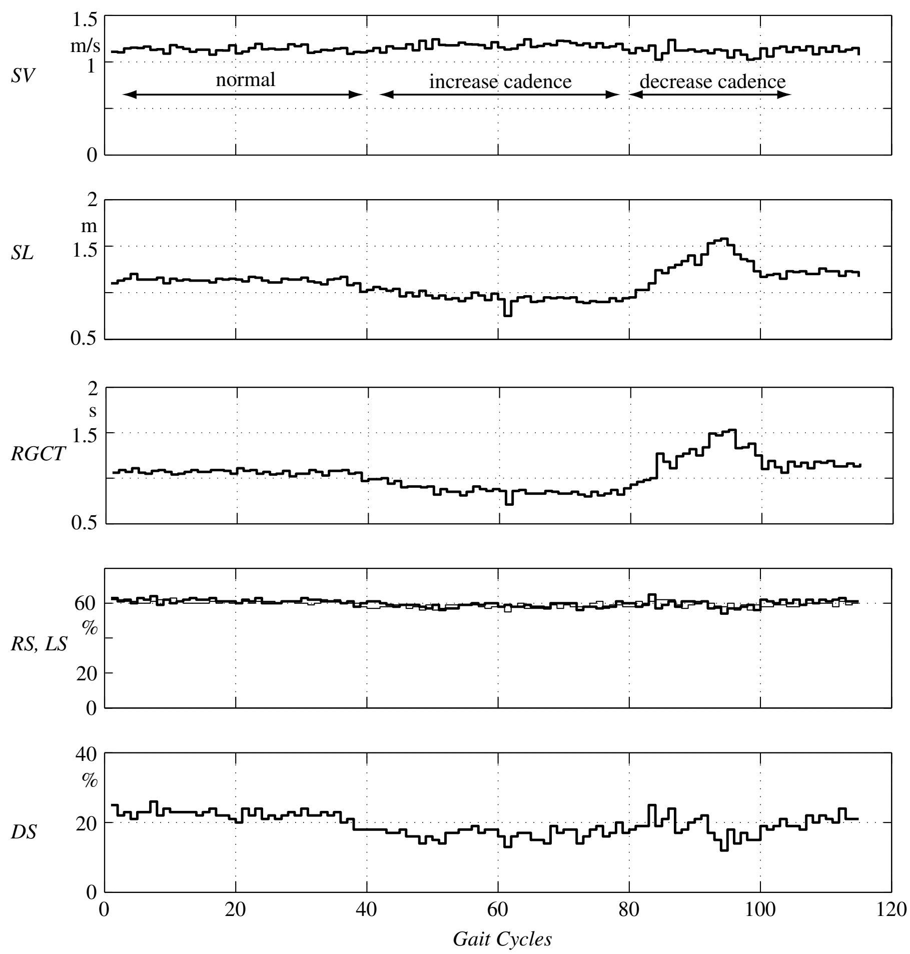

Fig. 10. Stride velocity (SV), stride length (SL), gait cycle time (RGCT), right and left stance (RS and LS) and double support (DS) in the percent of gait cycle obtained from a young subject during 2 min of treadmill walking using the proposed method. The velocity of treadmill was kept at 1.11 m/s while the subject changed his cadence. All parameters are estimated at each gait cycle allowing to analyse the stride to stride variability.

confidence interval of stance period equal to [−20 ms, 20 ms]. In the present study, which involve young and elderly subjects the error of toe-off and heelstrike detection is less that 5 ms (95%), with a systematic delay for heelstrike time of 10 ms.

Gait events can also be explained by looking at the particular pattern of the shank and thigh angular velocities (Fig. 5). Shank angular velocity (signal s_R) is a bipolar signal. Considering as positive the counter-clockwise rotation, one can observe that during the swing period s_R is rather positive while in stance phase s_R is almost negative. Just prior to heelstrike, s_R becomes negative. At heelstrike, there is a flexion (Vaughan et al., 1992) of the knee shown by a negative s_R and positive t_R. The time of heelstrike is also characterized by a change of slope sign of s_R. This is expected since before heelstrike, the shank decelerates (negative slope) to stop and after heelstrike there is an acceleration (positive slope) due to rotation around the ankle. After heelstrike, there is a clockwise rotation of the shank around the ankle involving a negative s_R. In mid-stance, the knee becomes straight leading to a slight angular velocity, which could become positive. Then the shank continues its clockwise rotation around the ankle and accelerates its rotation at the toe-off (positive slope of s_R). The rotation is still clockwise (s_R negative) at the beginning of the swing phase until the knee reaches its maximum flexion (t_R is positive and s_R negative), then the knee comes into extension (s_R positive and t_R positive) moving the shank from backward to forward. In this way, shank and thigh angular velocities also give remarkable insight into the recognition of the periods of knee flexion and extension during gait cycle.

In order to measure spatial parameters such as stride length, we have used the change of lower limb angles. Miyazaki (1997) has shown that for the estimation of the stride length from the integral of angular velocity, the relative change of angular rotation is sufficient and therefore there is no need of initial angular position. However this method consider thigh and shank as a unique segment and the error reached 15%. We propose a double segment gait model involving both shank and thigh, which consider a double pendulum model during the swing phase and the inverse double pendulum model during the stance phase. The identification of swing and stance phases was therefore necessary, and was determined by toe-off and heelstrike estimation. The estimation error for velocity and stride length is around 7%. This error is principally due to the inaccuracy of the gyroscopes which induce some error on α and β in (20) and (25). This error could increase if the period of integration increases considerably (very slow speed). In this case a new calibration of (29) and (30) is necessary. Other factors such as the difficulty to measure the exact shank and thigh length, the misalignment of the gyroscope axis regarding the mediolateral axis and the error due to gait phase detection are also to be considered. The low values of the intercept (close to 0) and the slope close to 1 in (29) and (30) confirm the appropriateness of the proposed gait model.

In this study, the stride length and velocity were estimated based on the right leg's angular velocity and assuming that in straight walking, right and left stride length are equal. If this assumption is no longer valid (e.g. curled trajectory or asymmetric gait) right and left leg rotation can be estimated separately by adding a 4th gyroscope on the left thigh. However, few of our elderly subjects had slightly asymmetric gait and we did not observe significant variation in the performance of the system. There are many advantages to use gyroscopes instead of other kinematic sensors such as accelerometers. First, unlike the accelerometer, the gyroscope can be attached to anywhere of any body segment as long as its axis is parallel to the mediolateral axis: the angular rotation is still the same along this segment. Tong and Granat (1999) have shown that the signals from different gyroscopes at different attachment site are almost identical. Second, the angular rate signal is less noisy than acceleration since acceleration is the derivative of velocity and involves higher frequency components. Third, accelerometry provides essentially temporal gait parameters, using singular peak enhancement. In contrast, rotation angles can be estimated from angular velocity by simple integration, and provides spatial gait parameters in addition to temporal parameters. Recently, the possibility to estimate knee angle (Williamson and Andrews, 2001) and trunk tilt (Najafi et al., 2001a) has also been shown. Finally, there is no influence of gravity acceleration on the measured signal. Gyroscopes however, do have some weaknesses. First, the piezoelectric gyroscope is more delicate to use than accelerometer. It is more sensitive to temperature and shock due to the mechanical fastening of vibrating piezo beam inside the sensor's case. In addition, powerful signal processing and filtering are necessary to cancel drift and artefact in the signal. Nevertheless, this filtering was done in this study by using wavelet transformation. The use of wavelet was particularly important to ensure a good time resolution in finding gait events such as toe-off and heelstrike.

Overall, the measuring device is light (300 g), portable, low cost and involves no discomfort for the patient who carries it for a long period of time. The datalogger is battery powered with a lifetime of 20 h. It can collect data up to 8 Mbytes corresponding to a period of 2.5 h (with three sensors at 200 Hz sampling rate). Additional memory cards or lower sampling rate can be used to increase the period of recording. The off-line calculation of temporal and spatial parameters is very fast: for a walking trial of 2 min, it takes from few seconds to less than a second, depending on computer's performance.

5. Conclusion

The proposed method appears a promising monitoring tool for several purposes. First, it allows measurements of gait features during a long period of walking and thus supplies the stride-to-stride variability of gait as illustrated in Fig. 10. In addition, because of its portability, this system can be used in other settings than a gait laboratory and therefore provides information that is more likely to reflect the actual performance of the subjects.

It can be used in many clinical applications such as outcome evaluation after knee and hip replacement, or external prosthesis adjustment for amputees. In elderly subjects, this system can also be proposed as a diagnostic tool for abnormal gait analysis, as a predictor tool for fall risk estimation (Najafi et al., 2001b), or as a monitoring tool to assess progress through rehabilitation. During this study the system proved to have a high acceptability by elderly subjects.

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