J. Biomechanics Vol. 23, No. 8, pp. 859 863, 1990.
Printed in Great Britain

0021 9290/90 $3.00 + .00
Pergamon Press plc

TECHNICAL NOTE

REAL-TIME GAIT ASSESSMENT UTILIZING A NEW WAY OF ACCELEROMETRY

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Abstract—Real-time registration of body segment angles is essential in artificial body position control. A new method is presented for the real-time calculation of the lower extremity angles using data obtained from pairs of two one-dimensional accelerometers. It is shown that, assuming rigid-body dynamics and simple hinge joints, relative angles (i.e. angles between segments) can be calculated without integration, thereby solving the problem of integration drift normally associated with accelerometry. During the stance phase of walking, the relative angles can be transformed to absolute angles (i.e. relative to the gravitational field direction) for the different leg segments. The feasibility of relative angle calculation is demonstrated by calculation of the knee angle of a healthy subject. Stability and resolution were demonstrated with measurements during standing. Measurements during standing up, sitting down and walking showed that shock (heel-strike) and skin movements, due to movements of the underlying muscle tissue, are the main error sources. Additional signal processing, e.g. low-pass filtering, can be used to diminish this error. The accuracy of the knee angle found is shown to be high enough to be used in a feedback controller for functional electrostimulation of the lower extremities.

NOMENCLATURE

(See Fig. 1.)

P_{ij}	point j of segment i
P_{i0}	origin of segment i
a_{ij}	equivalent acceleration in P_{ij}
F_{ij}	force working in P_{ij} on m
$F_{r_{ij}}$	reaction force working on m
g	gravitational acceleration
m	seismic mass of accelerometer
r_{ij}	position of P_{ij} relative to the body-fixed frame
x_{ij}, y_{ij}	inertial reference system coordinates of P_{ij}
θ_i	angle of segment i, relative to the vertical

The points P_{ij} are placed on the line from the joint between segment $i-1$ and i and the joint between segment i and $i+1$.

INTRODUCTION

Functional electrostimulation (FES) (Vodovnik *et al.*, 1981) of paralyzed muscles for regaining functional movements is hampered by problems resulting from, for example, fatigue and variability of the stimulation. These result in poor reproducibility of movements. A substantial increase in performance quality is expected from feedback control systems (Petrofsky and Phillips, 1986; McNeal and Bekey, 1984; Wilhere *et al.*, 1985). These should not only be used to increase reproducibility and 'quality' of walking, but also to lessen muscle fatigue. The quality achievable with a feedback system depends largely on the assessment of feedback data. For the lower extremities these are angle, angular velocity and angular acceleration of the ankle, knee and hip. Sensors for the assessment of feedback data should be small, lightweight, robust and have a low power consumption. Furthermore, a potential for implantation is desirable. To the best of the authors' knowledge, none of the present available angle sensors can fulfill all these criteria. However, in theory, accelerometers can. Therefore we studied their potential as sensors for the use in feedback-controlled FES systems.

Accelerometers have been used for the registration of body movements. However their use was restricted because the angle was calculated by integration of angular acceleration or angular velocity, and drift distorted the results. So accelerometers have been used for angular velocity and angular acceleration measurements only (Padgaonkar *et al.*, 1975) or for angle measurements during short intervals (Mital and King, 1979; Smidt *et al.*, 1977). Morris (1973) estimated the amount of drift by identifying the different walking cycles and by comparing the angle at the beginning and end of each cycle. This way drift can be removed. For real-time applications however, such as feedback control systems, this technique can not be used. In this study we show that, using pairs of accelerometers, relative angles (i.e. ankle, knee and hip angles) can be calculated without integration, thereby solving the problem of integration drift.

THEORY

A multi-segment body is moving in space. A seismic accelerometer capable of measuring static forces is placed in P_{ij} (Fig. 1). Considering the almost two-dimensional character of leg motion during locomotion we assume that a two-dimensional approach will be sufficient. In that case, the forces on the seismic mass, expressed in inertial reference coordinates, are given by Newton's law as:

$$\sum F_{ij}^x = F_{r_{ij}}^x = m\ddot{x}_{ij} \tag{1a}$$

$$\sum F_{ij}^y = F_{r_{ij}}^y - mg = m\ddot{y}_{ij}. \tag{1b}$$

The acceleration signal from the accelerometer including the gravitational component, called equivalent acceleration, is

Received in final form 20 December 1989.

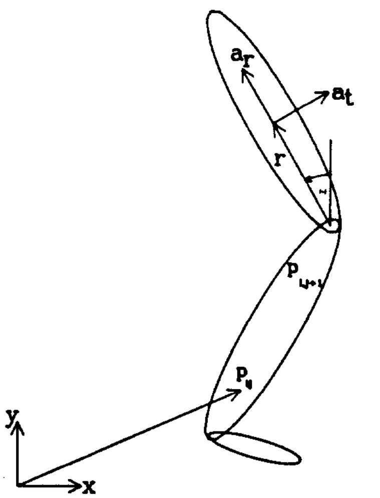

Fig. 1. Nomenclature for a multi-segment body with inertial reference and body-fixed frame.

given by:

$$a^x_{ij} = -F^x_{e,ij}/m = -\ddot{x}_{ij} \tag{2a}$$

$$a^y_{ij} = -F^y_{e,ij}/m = -\ddot{y}_{ij} - g. \tag{2b}$$

Writing x_{ij} and y_{ij} as

$$x_{ij} = x_{i0} - r_{ij}\sin(\theta_i) \tag{3a}$$

$$y_{ij} = y_{i0} + r_{ij}\cos(\theta_i), \tag{3b}$$

and assuming rigid body conditions and a hinge type knee joint, i.e. $\dot{r}_{ij} = \ddot{r}_{ij} = 0$, we find

$$\ddot{x}_{ij} = \ddot{x}_{i0} + r_{ij}\sin(\theta_i)\dot{\theta}_i^2 - r_{ij}\cos(\theta_i)\ddot{\theta}_i \tag{4a}$$

$$\ddot{y}_{ij} = \ddot{y}_{i0} - r_{ij}\cos(\theta_i)\dot{\theta}_i^2 - r_{ij}\sin(\theta_i)\ddot{\theta}_i. \tag{4b}$$

Because the accelerometers are attached to the leg, we measure body-fixed accelerations. These can be found by multiplication of the accelerations, expressed in inertial reference coordinates, with a rotation matrix.

$$\begin{aligned} a^t_{ij} &= \cos(\theta_i)a^x_{ij} + \sin(\theta_i)a^y_{ij} \\ &= -g\sin(\theta_i) - (\cos(\theta_i)\ddot{x}_{i0} + \sin(\theta_i)\ddot{y}_{i0}) + r_{ij}\ddot{\theta}_i \end{aligned} \tag{5a}$$

$$\begin{aligned} a^r_{ij} &= -\sin(\theta_i)a^x_{ij} + \cos(\theta_i)a^y_{ij} \\ &= -g\cos(\theta_i) - (-\sin(\theta_i)\ddot{x}_{i0} + \cos(\theta_i)\ddot{y}_{i0}) + r_{ij}\dot{\theta}_i^2. \end{aligned} \tag{5b}$$

The right-hand side of equation (5) comprises three acceleration contributions. The first describes the contribution of the gravitational field. The second represents the accelerations due to the body-fixed frame translation, whereas the last part represents the acceleration due to the body-fixed frame rotation.

Placing the body-fixed frames of segments i and $i+1$ at the joint connecting the two segments, we have $P_{i0} \equiv P_{i+1,0}$. Consequently,

$$a^x_{i0} = a^x_{i+1,0} \tag{6a}$$

$$a^y_{i0} = a^y_{i+1,0}. \tag{6b}$$

Using the inverse rotation matrix this can be written as

$$\begin{aligned} \cos(\theta_i)a^t_{i0} - \sin(\theta_i)a^r_{i0} &= \cos(\theta_{i+1})a^t_{i+1,0} \\ &\quad - \sin(\theta_{i+1})a^r_{i+1,0} \end{aligned} \tag{7a}$$

$$\begin{aligned} \sin(\theta_i)a^t_{i0} + \cos(\theta_i)a^r_{i0} &= \sin(\theta_{i+1})a^t_{i+1,0} \\ &\quad + \cos(\theta_{i+1})a^r_{i+1,0} \end{aligned} \tag{7b}$$

where a^t_{i0} and a^r_{i0} can be obtained from equation (5) by placing accelerometers at P_{ij} and $P_{i,j+1}$, resulting in

$$a^t_{i0} = \frac{r_{i,j+1}a^t_{ij} - r_{ij}a^t_{i,j+1}}{r_{i,j+1} - r_{ij}} \quad a^r_{i0} = \frac{r_{i,j+1}a^r_{ij} - r_{ij}a^r_{i,j+1}}{r_{i,j+1} - r_{ij}}. \tag{8}$$

Rewriting equation (7) as

$$a^t_{i+1,0} = \cos(\theta_{i+1} - \theta_i)a^t_{i0} + \sin(\theta_{i+1} - \theta_i)a^r_{i0} \tag{9a}$$

$$a^r_{i+1,0} = -\sin(\theta_{i+1} - \theta_i)a^t_{i0} + \cos(\theta_{i+1} - \theta_i)a^r_{i0} \tag{9b}$$

we find

$$\tan(\theta_{i+1} - \theta_i) = \frac{a^t_{i+1,0}\,a^r_{i0} - a^r_{i+1,0}\,a^t_{i0}}{a^t_{i+1,0}\,a^t_{i0} + a^r_{i+1,0}\,a^r_{i0}}. \tag{10}$$

So, by comparing the (body-fixed) equivalent accelerations at the connecting joint we can calculate the joint angle, provided that the movements can be considered to be two-dimensional and that the distances between sensors and the joint are constant. The advantage of this method is the avoidance of integration.

During the stance phase of walking, $\ddot{x}_{ankle} \approx \ddot{y}_{ankle} \approx 0$, and equation (8) reduces to

$$a^t_{shank,0} = -g\sin(\theta_{shank}) \tag{11a}$$

$$a^r_{shank,0} = -g\cos(\theta_{shank}). \tag{11b}$$

So during the stance phase of walking, θ_{shank} can be calculated with accelerometers on the shank only, and is relative to the gravitational field direction and not to the ground.

METHODS

To calculate the knee angle by application of equation (10), eight uniaxial accelerometers (Kyowa AS-5GA) were used to measure the radial and tangential accelerations of the thigh and calf segments in the sagittal plane. The accelerometers were attached in pairs on PVC brackets, which were then attached to the leg with VELCRO straps. Two pairs were mounted on the lower leg bracket and two pairs on the upper leg bracket. A flexible goniometer permitting free movement (Penny + Giles G180) was mounted between the two brackets to measure the knee angle in the sagittal plane (Fig. 2). The lower leg bracket was oriented between ankle and knee joint and the upper leg bracket between knee and hip joint. All signals were amplified and low-pass filtered at 100 Hz. The signals were sampled at 500 Hz on a LSI-11/23 computer for

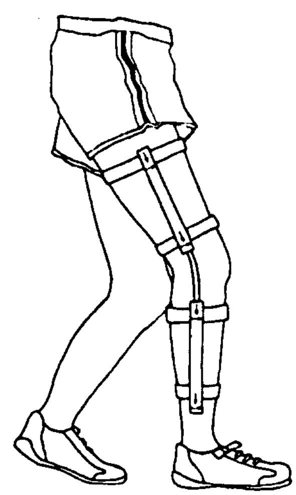

Fig. 2. Fixation of accelerometers and flexible goniometer to the lower extremities.

4 s. A total of 50 measurements comprising five different kinds of movements were analyzed for this study. Measurements were made during standing, standing up, sitting down, walking and while sitting (pendulum measurements), of a healthy subject wearing normal footwear.

RESULTS

The knee angle was calculated from the accelerometer data using equation (10). This was then compared with the knee angle as measured with the flexible goniometer. We found an average error (± 0.05 rad) which was compensated for. Of the remaining error we calculated the standard deviation using

$$\sigma^2 = \frac{1}{n} \sum_{t=0}^{n} (\theta_{gt} - \theta_{at})^2 \tag{12}$$

with σ = standard deviation

θ_{gt} = flexible goniometer knee angle

θ_{at} = accelerometer knee angle

t = time index.

For the pendulum measurements, the subject was sitting on a bench, the thigh being supported by the bench. The shank and knee could move freely. A typical result is shown in Fig. 3A. The standard deviations for the difference between calculated and measured knee angle ranged from 0.007 to 0.040 rad. For standing up (Fig. 3B), as well as for sitting down (Fig. 3C), standard deviations ranged from 0.03 to 0.06 rad. A typical result for the knee angle during walking is shown in Fig. 4A. The calculated knee angle shows large, high frequency distortions. The frequency of the oscillations at heel-strike depends on footwear and ranges from 5 to

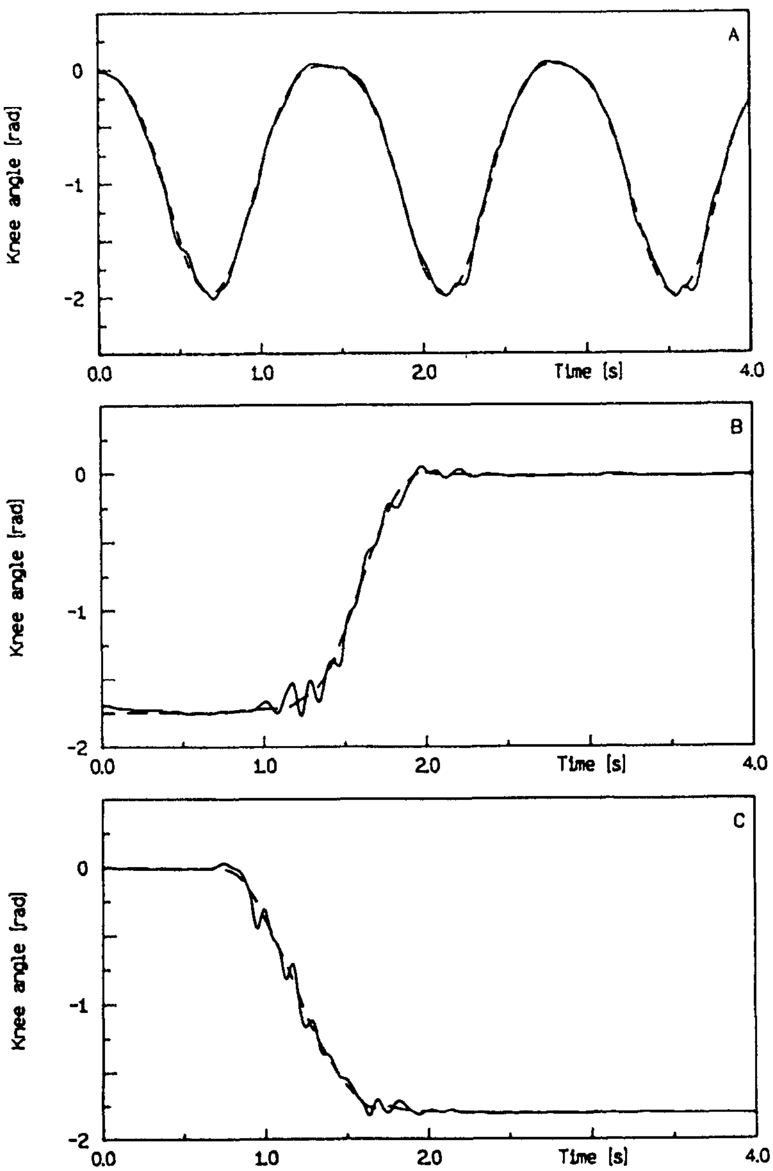

Fig. 3. Knee angle as calculated from accelerometer data (solid line) and as measured with a flexible goniometer line (broken line) as a function of time. (A) Shank pendulum measurement; (B) standing up; (C) sitting down.

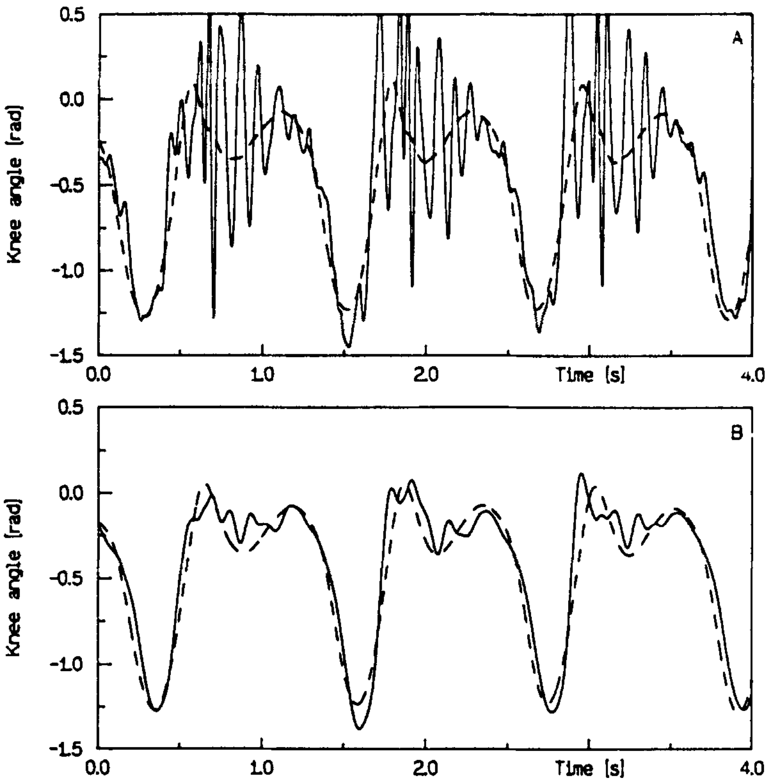

Fig. 4. Knee angle as calculated from accelerometer data (solid line) and as measured with a flexible goniometer line (broken line) as a function of time during walking. (A) Without filtering; (B) after 5 Hz filtering.

Table 1. Standard deviation for the knee angle as calculated from accelerometer data, compared to the knee angle as measured with a flexible goniometer. Low values are for slow movements, high values for fast movements

Movement	No filtering	After 5 Hz filtering
Standing	0.003	—
Pendulum	0.007–0.040	0–0.03
Standing up	0.03–0.06	0.02–0.04
Sitting down	0.03–0.06	0.02–0.05
Walking	0.1–0.2	0.04–0.09

20 Hz. Without additional signal processing, this would give standard deviations up to 0.2 rad. This can be reduced by low-pass filtering as is demonstrated by using a digital fourth-order low-pass Butterworth filter with a cut-off frequency of 5 Hz. The filtered knee angles are shown in Fig. 4B. The standard deviation after filtering both the measured and the calculated knee angle, ranges from 0.04 to 0.09 radians. High frequency components in the calculated knee angle are apparent for standing up and sitting down also. Additional filtering reduced the standard deviations by 10–20%. The results for the different measurements are summarized in Table 1.

DISCUSSION

Various methods have been described (Gilbert *et al.*, 1984; Hayes *et al.*, 1983; Morris, 1973) to obtain the position of the leg without integration during the stance phase of walking using accelerometer data. The leg position during the whole walking cycle, however, is obtained by integration of the angular acceleration or angular velocity. This introduces integration drift. Because of the cyclic nature of human walking, the drift can be estimated. Afterwards it can then be compensated for. For real-time applications, however, such as feedback control systems, this cannot be used. As shown, using pairs of accelerometers, the knee angle can be calculated without integration, thereby solving the problem of integration drift. During stance, one can also calculate the absolute segment angle which might be beneficial for the closed-loop control of standing, especially concerning the problem of balance.

Comparing the knee angle as calculated from accelerometer data, and as measured with the flexible goniometer, an average error was found. In experiments with standing and walking using feedback controlled FES, the knee angle during erect standing is usually defined as zero. So, average errors can be neglected. The calculated knee angle during standing up shows an error before the movement actually starts. This was caused by movements of the quadriceps muscles when building up force. These movements of the underlying muscles are detected by the accelerometers and interpreted as a change in the knee angle. A related effect can be seen during sitting down, where the movement is controlled by increasing quadriceps muscle force. This stresses the importance of fixation of the accelerometers. Implantation of the accelerometers, with fixation to the bones, may solve this problem but is unlikely in the near future. In the mean time, however, the use of accelerometers in a hybrid system with fixation to the orthosis could be studied.

The fixation problem is even more pronounced during walking. The high-frequency errors can be reduced by additional low-pass filtering as was demonstrated. In effect, the frequency of the heel-strike related error then sets an upper limit to the bandwidth of the knee angle signal and thereby to the bandwidth of the FES controller. Theoretically, however, the high accelerations occurring at heel-strike should have no effect on the calculated knee angle, contrary to the results presented in Fig. 4. The frequency of the accelerations at heel-strike was found to depend on the type of footwear, and ranged from 5 to 20 Hz. So the resonance frequency of the muscle–accelerometer system must be at least 20 Hz, indicating a good fixation of the brace system. Applying a low-pass filter with a cut-off frequency below 20 Hz can thus ensure that the results are not affected by potential resonances of the muscle accelerometer system. Lower frequency effects, such as movements of muscle tissue, can still affect the results, as was shown for standing up and sitting down. The error in the knee angle can be caused also by the non-ideal knee joint or by not fulfilling the rigid-body condition or the two-dimensional approximation. Additional measurements, with the combined measurement of positions and accelerations, are needed to establish the influence of these errors. Remaining error sources are the flexible goniometer [errors of 0.02 rad for standard goniometers were reported by Chao (1980)] and measurement artifacts. Due to the high angular velocity during the swing phase of walking (± 8 rad s^{-1}), even small time delays between the two knee angle signals (e.g. caused by the consecutive analog to digital conversion or by differences in the low-pass filters) can lead to relatively large errors.

The errors given are a function of speed. Because the movements of a paraplegic walking with FES are usually slow, we expect the lower range to be a good estimate of the results that can be achieved with these patients. Preliminary data collected by Crago *et al.* (1986) indicate errors of 0.015 rad (standing) and 0.05 rad (walking) to be tolerable for closed-loop control of the knee joint by FES, which is then comparable with the results we found. We conclude that the assumptions made (rigid-body dynamics, two-dimensional movements and a hinge knee joint) do not lead to unacceptable errors and that our method for calculating the knee angle from accelerometer data can be used in feedback controlled FES systems.

The results presented here are a first step towards the development of an implantable goniometer based on accelerometers. A number of both technical and medical problems will have to be solved before implantation can be considered. For instance, on the technical side the development of specific miniature accelerometers e.g. using Si-technology (Chen *et al.*, 1982; Roylance and Angell, 1979), preferably with integrated electronics (Petersen *et al.*, 1982), is necessary. On the medical side, the need to prevent movement of the accelerometers will ask for fixation to the bones. Problems concerning the encapsulation of the sensors and the leads will have to be addressed also.

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