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Short communication

Walking speed estimation using a shank-mounted inertial measurement unit

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ABSTRACT

We studied the feasibility of estimating walking speed using a shank-mounted inertial measurement unit. Our approach took advantage of the inverted pendulum-like behavior of the stance leg during walking to identify a new method for dividing up walking into individual stride cycles and estimating the initial conditions for the direct integration of the accelerometer and gyroscope signals. To test its accuracy, we compared speed estimates to known values during walking overground and on a treadmill. The speed estimation method worked well across treadmill speeds and slopes yielding a root mean square speed estimation error of only 7%. It also worked well during overground walking with a 4% error in the estimated travel distance. This accuracy is comparable to that achieved from foot-mounted sensors, providing an alternative in sensor positioning for walking speed estimation. Shank mounted sensors may be of great benefit for estimating speed in walking with abnormal foot motion and for the embedded control of knee-mounted devices such as prostheses and energy harvesters.

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1. Introduction

Portable sensors for estimating walking speed are potentially useful for diagnosing abnormal gait, evaluating the effectiveness of rehabilitation techniques, and controlling biomedical devices. Walking speed, cannot be determined by simply integrating signals from portable accelerometers. The first complication is that measured accelerations have contributions not only from the motion of the limb but also from gravity. Gravitational acceleration must be subtracted before integration requiring continuous knowledge of the device angle with respect to gravity. Second, accelerometer drift can quickly cause large inaccuracies in the estimated displacement because its contribution grows proportional to the square of time. A solution is to take advantage of the cyclical nature of walking and divide up the continuous motion into a series of stride cycles, resetting integration at the beginning of each new cycle. Resetting treats the new offset as a constant and as long as cycle duration is much shorter than the rate at which the accelerometer offset is changing, quite accurate results can be achieved. However, resetting introduces a third complication—the initial condition for each integration must be known. Sabatini and colleagues were able to account for these complications and directly integrate measured accelerations from a foot-mounted inertial measurement unit—a sensor that combines accelerometers and gyroscopes—to yield accurate estimates of

walking speed (Sabatini et al., 2005). Their method took advantage of the unique mechanics of the foot during locomotion—that it has zero linear and angular velocity once per stride when the foot is flat on the ground—to determine the initial conditions for each new integration.

The purpose of this paper was to study the feasibility of estimating walking speed using a shank-mounted inertial measurement unit (IMU). Mounting sensors close to the knee would be useful for the embedded control of prostheses, and energy harvesters (Herr and Wilkenfeld, 2003; Donelan et al., 2008). In addition, it is not always desirable to mount sensors directly on the feet as they may move out of the plane of progression during pathological gait (Jasiewicz et al., 2006). Our approach took advantage of the inverted pendulum-like behavior of the stance leg during walking to identify a new method for segmenting the gait cycle and estimating the initial conditions for integration.

2. Methods

2.1. Speed estimation

Shank linear accelerations and angular velocity were measured using a bi-axial accelerometer (Analog Devices ADXL320) and a gyroscope (Analog Devices ADXRS300), respectively. To compute the displacements along the horizontal and vertical world coordinate axes, $a_x(t)$ and $a_y(t)$ at time t into component accelerations $a_x(t)$ and $a_y(t)$ in the world coordinate system (Fig. 1) according to

$$\begin{bmatrix} a_x(t) \\ a_y(t) \end{bmatrix} = \begin{bmatrix} \cos\theta(t) & -\sin\theta(t) \\ \sin\theta(t) & \cos\theta(t) \end{bmatrix} \begin{bmatrix} a_t(t) \\ a_n(t) \end{bmatrix} - \begin{bmatrix} 0 \\ g \end{bmatrix}, \quad (1)$$

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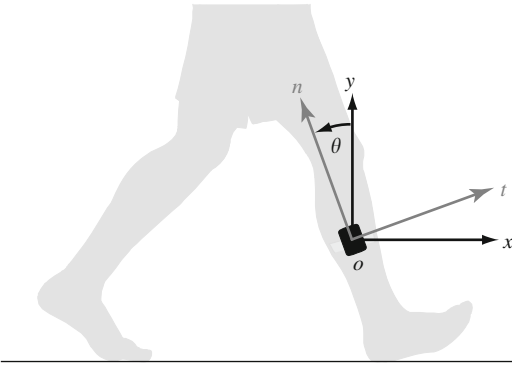


Fig. 1. We attached an inertial measurement unit to the lateral-aspect of each subject's shank. The normal and tangential accelerations were measured along the n and t axes, respectively. The gyroscope axis is orthogonal to the plane defined by the tangential and normal axes. We defined the world coordinate as xoy , with the vertical axis y parallel to gravity. Shank angle, θ , is defined as the angle between the normal axis of the accelerometer and the vertical axis of the world coordinate system. As per the right hand rule, positive angular velocities correspond to a counterclockwise rotation of the shank. Arrows indicate positive directions for each axis.

where g is the acceleration due to the gravity and $\theta(t)$ is the shank angle which was computed by integrating the measured angular velocity $\omega(t)$,

$$\theta(t) = \int_0^t \omega(\tau) d\tau + \theta(0), \quad (2)$$

where $\theta(0)$ is the initial shank angle before integration.

We segmented the continuous walking motion into a series of stride cycles before computing the displacements. Mid-stance shank vertical events—the time in the stance phase when the shank is parallel to the direction of gravity—defined each new stride cycle (Fig. 2). The inverted pendulum-like behavior of the stance leg during walking allowed us to identify each mid-stance vertical event from a characteristic feature in the gyroscope signal (Fig. 2). By definition, $\theta(0) = 0$ at mid-stance vertical providing the initial condition for integrating Eq. (2). Moreover, the initial horizontal velocity $v_x(0)$ and vertical velocity $v_y(0)$ are approximately zero because of the inverted-pendulum behavior. These initial conditions allowed to compute the displacements over each stride cycle duration, T .

Within each stride cycle $[0, T]$, we computed the associated velocities $v_x(t)$ and $v_y(t)$,

$$v_x(t) = \int_0^t a_x(\tau) d\tau + v_x(0),$$

$$v_y(t) = \int_0^t a_y(\tau) d\tau + v_y(0), \quad (3)$$

where $v_x(0) = 0$ and $v_y(0) = 0$.

By integrating the velocities $v_x(t)$ and $v_y(t)$, we obtained the horizontal displacement, $s_x(T)$, and vertical displacement, $s_y(T)$,

$$s_x(T) = \int_0^T v_x(\tau) d\tau - \frac{1}{2}T \cdot v_x(T),$$

$$s_y(T) = \int_0^T v_y(\tau) d\tau - \frac{1}{2}T \cdot v_y(T), \quad (4)$$

where $v_x(T)$ and $v_y(T)$ are the horizontal and vertical velocities at the end of the stride. The second term is a correction term for reducing the displacement estimation errors associated with the acceleration offsets.

At the end of each stride cycle, we computed the stride length s_T , slope $\phi(T)$, in percent grade, and the average walking speed $V(T)$, in m/s, as

$$s_T = \sqrt{s_x^2(T) + s_y^2(T)},$$

$$\phi(T) = \frac{s_y(T)}{s_x(T)} \cdot 100,$$

$$V(T) = s_T/T. \quad (5)$$

The walking distance s_d was computed by summation of the estimated individual stride lengths as

$$s_d = \sum_{n=0}^N s_T(n). \quad (6)$$

where N is the total number of strides.

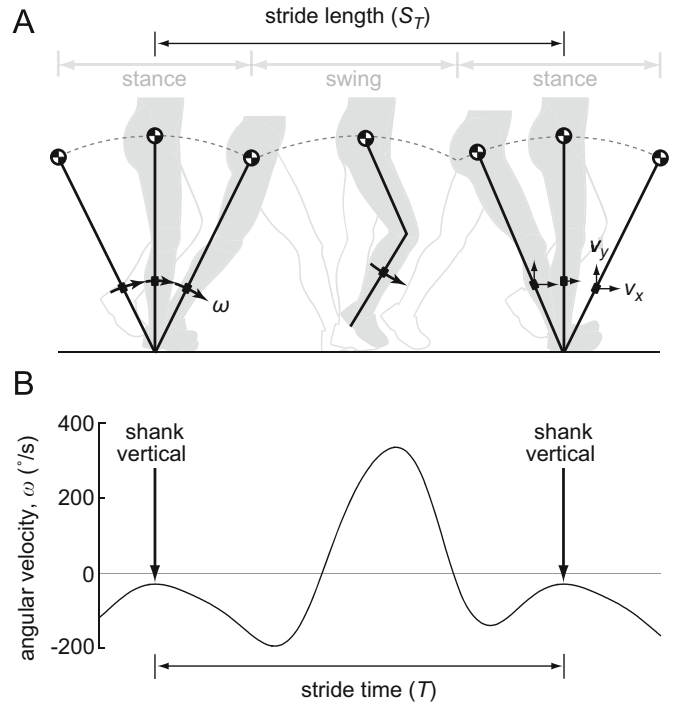


Fig. 2. During the inverted pendulum-like stance phase (A), the body vaults up and over the stance leg with measured shank angular velocity (B) negative and slowing down as kinetic energy is exchanged for potential energy (Cavagna et al., 1977). At the shank vertical event, the body COM reaches its highest point, potential energy reaches a maximum, and velocity reaches a minimum. Consequently, the IMU horizontal and vertical velocities (v_x and v_y) are approximately zero at mid-stance. The angular velocity of the shank (ω) is slowest at this point but then accelerates as the inverted pendulum accelerates downward, exchanging potential energy for kinetic energy. The angular velocity of the shank switches from negative to positive during swing in order to progress the shank forward and return it to the correct orientation at the beginning of the next stance phase. Stride length, S_T , and stride time, T are defined as the distance and time between two mid-stance shank vertical events. The characteristic feature for defining mid-stance vertical events was the local maximum during the lengthy period of negative shank angular velocity.

Before each walking experiment, we calibrated the accelerometer by aligning its axes parallel and perpendicular to gravity (nominal output of 1 and 0g, respectively) and adjusted the gain and offset accordingly. We performed a single calibration for the gyroscope using a dynamometer (BIODEX II, Biodex Medical Systems, New York, $\pm 1^\circ/s$ accuracy) to rotate the device at angular velocities from 30 to 300 $^\circ/s$ with 30 $^\circ/s$ increments. As a result of the calibration, we set the sensor bias to 2.53 V (the nominal value was 2.5) and the gain to 0.0048 V/ $^\circ/s$ (the nominal value was 0.005). The algorithm was programmed in Simulink, and executed at 1 KHz using Real Time Windows Target (Mathworks, Natick, MA).

2.2. Experimental methods

Five male and three female subjects (age: 28.0 ± 5.8 years; height: 1.68 ± 0.07 ; tibia length: 0.41 ± 0.03 m) participated in the treadmill walking experiment. The three female subjects also performed overground walking experiments. Before the experiments began, volunteers gave their informed consent to participate in accordance with university policy. We collected data at treadmill speeds of 0.8, 1.0, 1.2, 1.4, 1.6, and 1.8 m/s. At each speed, subjects walked at -10% , -5% , 0% , 5% , and 10% grade. During each 90 s trials, an IMU was attached with athletic tape onto the shank in the sagittal plane with the center of IMU positioned midway between the knee and ankle. During the overground experiments, subjects completed two trials of walking along a straight 100 m long course at their preferred walking speed.

2.3. Data analysis

For each treadmill walking trial, we calculated the mean walking speed by averaging the stride-by-stride data from the last 60 s of each trial. Because shorter duration averages are sometimes desirable, we tested for a difference between a 20 and 60 s average and found no significant difference. Estimation error at a given speed and slope was calculated as the difference between the estimated speed and

the actual treadmill value. Within a condition, we averaged across subjects to determine the mean estimation error (mean) and standard deviation (S.D.). We also calculated the root mean square error (RMSE) of the speed estimates as $RMSE = \sqrt{\sum (\text{estimated} - \text{actual})^2 / N}$ (N is the number of samples). The effects of walking speed and slope on the absolute difference between estimated and actual speed estimation error were tested using repeated-measures ANOVA, with $P < 0.05$ considered statistically significant. For overground walking trials, we computed the mean and RMSE of the estimated walking distance.

3. Results

The proposed speed estimation method accurately estimated walking speed. Fig. 3 presents typical data from a single subject during level walking. For this representative subject, speed is estimated accurately and with low variability. This pattern holds across the eight measured subjects as summarized in Fig. 4 and Table 1. The algorithm tended to slightly overestimate at slow speeds and underestimate at fast speeds ($P=0.06$, $F=3.57$, $d.o.f.=47$). We did not find any apparent correlation between speed estimation error and subject-specific parameters such as height, weight, or shank length. Different subjects were responsible for the largest estimation errors at each speed. While walking slope had a systematic effect on estimated speed ($P=0.03$, $F=4.36$, $d.o.f.=39$), the speed estimation algorithm was generally accurate

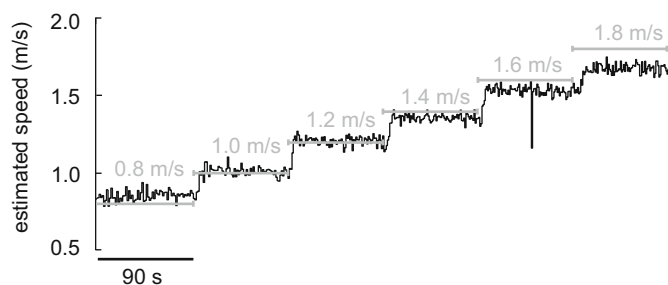


Fig. 3. Estimated speed from a representative subject during level walking at treadmill speeds ranging from 0.8 to 1.8 m/s.

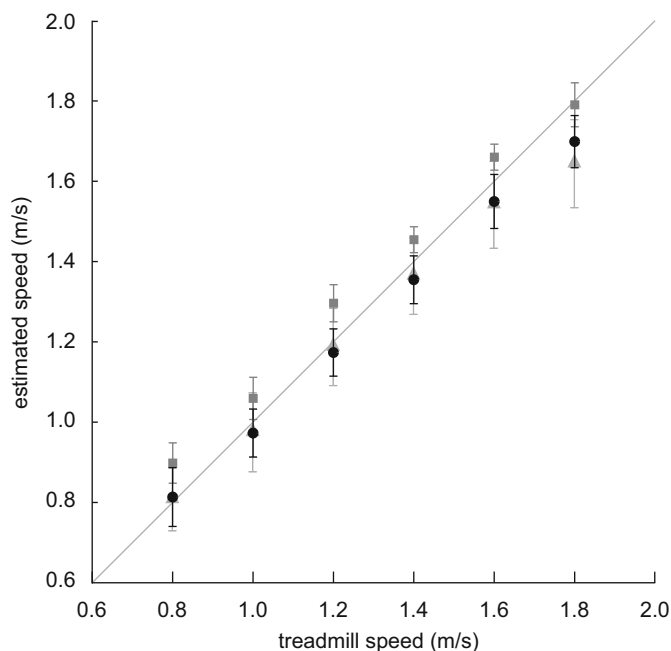


Fig. 4. Estimated speeds during -10% (\blacktriangle), 0% (\bullet) and 10% (\blacksquare) grade walking. The solid gray line is the line of identity where the estimated speed equals the treadmill speed. Values shown are means \pm S.D., $N = 8$.

across both speeds and slopes—the root mean square speed estimation error was only 7%. The accuracy of speed estimation resulted in accurate distance estimation. The mean value of the estimated distance covered in the six overground walking trials was 96.5 ± 2.0 m equating to a RMSE of 4%.

4. Discussion

While accurate for estimating walking speed, our algorithm failed to provide accurate slope estimation by consistently underestimating both positive and negative slopes. For example, at the steepest decline of -10% grade and the highest walking speed of 1.8 m/s, the algorithm estimated a positive slope of 5% grade. These inaccuracies appear to arise, at least in part, from limitations of setting the initial conditions (Eq. (2)) based on the predicted mid-stance shank vertical event. During incline walking, the COM kinetic energy reaches its minimum prior to the COM gravitational potential energy reaching its maximum (Gottschall and Kram, 2006). In the context of our method, this implies that the local maximum of shank angular velocity occurs prior to the stance leg reaching its vertical position during incline walking, corresponding to a positive initial shank angle. The order of the COM energy extrema is reversed during decline walking (Gottschall and Kram, 2006) suggesting that the local maximum of shank angular velocity occurs after the stance leg reaching its vertical position, corresponding to a negative initial shank angle. As we assumed a zero sensor angle at mid-stance shank vertical event (Eq. (2)), any deviation of the actual initial angle from zero would result in the same amount of offset in the shank angle. This offset is equivalent to a rotation of the local sensor coordinate frame—a redistribution of the normal and tangential acceleration (Eq. (1)) in the world coordinate system. Consequently, our method underestimated the vertical displacement for both incline and decline walking, and the slope as well. However, as observed in the experimental results (Table 1), the offset in the estimated shank angle had no measurable effect on speed estimation because the isometric transformation induced by rotation did not change the amplitude of the acceleration, and therefore the stride length (Eq. (5)). The inaccuracy in estimating slope highlights the method's dependence on the inverted pendulum-like behavior of the shank during stance. Consequently, the applicability of this method to speed estimation to running, or pathological gaits, will need to be tested empirically.

The measured errors in estimating speed is most likely due to the zero velocity assumption at mid-stance shank vertical (Eq. (3)). Any deviation of the actual initial horizontal velocity from zero would result in the same amount of offset in the estimated speed. Because the shank rotates about the ankle joint at the mid-stance shank vertical event, the absolute value of the initial horizontal velocity $v_x(0)$ is approximately equal to the product of the angular velocity ω of the shank and the distance of the sensor to the ankle joint (Fig. 2). At the mid-stance shank vertical event, the shank angular velocity reached a non-zero local maximum resulting in a positive non-zero initial horizontal velocity (Fig. 2). The speed estimation algorithm underestimated walking speed, and the underestimation became larger at faster walking speeds, because the shank retained a greater angular velocity, and therefore a larger initial horizontal velocity, at the peak of the inverted pendulum arc. One possible remedy is to estimate the initial velocity based on the angular velocity, and therefore achieve a higher accuracy.

The present method yields speed estimate accuracies roughly comparable to that achieved by Sabatini and colleagues who used a sensor mounted to the foot (Sabatini et al., 2005). That accurate results can be achieved with either method allows for flexibility in sensor positioning. This may be of great benefit for integrating

Table 1
Speed estimation error at different speed and slope.

| Slope | | Speed (m/s) | | | | | | RMSE |
|-------|------|-------------|--------------|--------------|--------------|--------------|--------------|------|
| | | 0.8 | 1.0 | 1.2 | 1.4 | 1.6 | 1.8 | |
| –10% | abs. | 0.01 ± 0.08 | –0.03 ± 0.10 | –0.01 ± 0.10 | –0.04 ± 0.10 | –0.06 ± 0.11 | –0.16 ± 0.11 | 0.12 |
| | pct. | 0.01 ± 0.1 | –0.03 ± 0.1 | –0.01 ± 0.08 | –0.03 ± 0.07 | –0.03 ± 0.07 | –0.09 ± 0.06 | 0.09 |
| –5% | abs. | 0.04 ± 0.07 | 0 ± 0.06 | 0.04 ± 0.06 | 0 ± 0.07 | –0.02 ± 0.07 | –0.09 ± 0.08 | 0.08 |
| | pct. | 0.05 ± 0.09 | 0 ± 0.06 | 0.03 ± 0.05 | 0 ± 0.05 | –0.02 ± 0.05 | –0.05 ± 0.04 | 0.06 |
| 0% | abs. | 0.01 ± 0.07 | –0.03 ± 0.06 | –0.03 ± 0.06 | –0.05 ± 0.06 | –0.05 ± 0.07 | –0.10 ± 0.07 | 0.08 |
| | pct. | 0.02 ± 0.09 | –0.03 ± 0.06 | –0.03 ± 0.05 | –0.03 ± 0.04 | –0.03 ± 0.04 | –0.06 ± 0.04 | 0.06 |
| 5% | abs. | 0.02 ± 0.06 | –0.02 ± 0.03 | 0.01 ± 0.03 | 0.02 ± 0.05 | –0.01 ± 0.04 | –0.06 ± 0.04 | 0.05 |
| | pct. | 0.02 ± 0.08 | –0.02 ± 0.03 | 0.01 ± 0.02 | –0.01 ± 0.03 | 0 ± 0.02 | –0.03 ± 0.02 | 0.04 |
| 10% | abs. | 0.10 ± 0.05 | 0.06 ± 0.05 | 0.10 ± 0.05 | 0.05 ± 0.03 | 0.06 ± 0.03 | –0.01 ± 0.05 | 0.08 |
| | pct. | 0.12 ± 0.06 | 0.06 ± 0.05 | 0.08 ± 0.04 | 0.04 ± 0.02 | 0.04 ± 0.02 | –0.01 ± 0.03 | 0.08 |
| RMSE | abs. | 0.08 | 0.07 | 0.08 | 0.07 | 0.08 | 0.12 | 0.08 |
| | pct. | 0.10 | 0.07 | 0.06 | 0.05 | 0.05 | 0.07 | 0.07 |
| Range | min. | 0.69 | 0.82 | 1.04 | 1.22 | 1.36 | 1.42 | |
| | max. | 0.98 | 1.14 | 1.38 | 1.54 | 1.71 | 1.85 | |

Absolute estimation error (abs.), percentage estimation error (pct.) at different speeds and slopes are shown. Values are means ± S.D., N=8.

sensors directly into shank or knee-mounted devices or for estimating speed in walking with abnormal foot motion.

Conflict of interest statement

Q.L., V.N. and J.M.D. have equity interest in Bionic Power Incorporated, a company that performs research and development on the energy harvesting technology. J.M.D. is chief science officer and board member of Bionic Power Incorporated.

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