Measurement of the mechanical power of walking by satellite positioning system (GPS)

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ABSTRACT

TERRIER, P., Q. LADETTO, B. MERMINOD, and Y. SCHUTZ. Measurement of the mechanical power of walking by satellite positioning system (GPS). *Med. Sci. Sports Exerc.*, Vol. 33, No. 11, 2001, pp. 1912–1918. **Purpose:** This descriptive article illustrates the application of Global Positioning System (GPS) professional receivers in the field of locomotion studies. The technological challenge was to assess the external mechanical work in outdoor walking. **Methods:** Five subjects walked five times during 5 min on an athletic track at different imposed stride frequency (from 70–130 steps·min^{-1}). A differential GPS system (carrier phase analysis) measured the variation of the position of the trunk at 5 Hz. A portable indirect calorimeter recorded breath-by-breath energy expenditure. **Results:** For a walking speed of 1.05 ± 0.11 m·s^{-1}, the vertical lift of the trunk (43 ± 14 mm) induced a power of 46.0 ± 20.4 W. The average speed variation per step (0.15 ± 0.03 m·s^{-1}) produced a kinetic power of 16.9 ± 7.2 W. As compared with commonly admitted values, the energy exchange (recovery) between the two energy components was low (39.1 ± 10.0%), which induced an overestimated mechanical power (38.9 ± 18.3 W or 0.60 W·kg^{-1} body mass) and a high net mechanical efficiency (26.9 ± 5.8%). **Conclusion:** We assumed that the cause of the overestimation was an unwanted oscillation of the GPS antenna. It is concluded that GPS (in phase mode) is now able to record small body movements during human locomotion, and constitutes a promising tool for gait analysis of outdoor unrestrained walking. However, the design of the receiver and the antenna must be adapted to human experiments and a thorough validation study remains to be conducted. **Key Words:** DIFFERENTIAL GLOBAL POSITIONING SYSTEM, ENERGY EXPENDITURE, INDIRECT CALORIMETRY, MECHANICAL EFFICIENCY

Human beings spend a substantial amount of time in walking, either for their work or for their leisure activities (1,5). Human locomotion has therefore been extensively studied in terms of energetics or biomechanics (6–10,12,17,25,26). Most of the classical studies have been realized under laboratory conditions. The typical analytical instruments have been force-plates or video recordings coupled, if required, with indirect calorimetry (17). Such classical techniques have permitted a better understanding of the walking mechanisms (6,7,9). However, one drawback is inherent to laboratory experiments: 3D video and force plate analyses fail to be deployed in a large unlimited space. Because of the limited experimental perimeter, it is not possible to record a substantial number of free consecutive steps. In contrast, treadmills allow one to mimic normal gait with good accuracy, but this technique may slightly modify the gait style under certain circumstances. Moreover, in a real-life situation, the subjects freely select a specific pathway to travel from one point to another. Such behavior is difficult to reproduce on a treadmill or analyze in a closed space. Consequently, we postulated that a new tool able to analyze walking outside the lab would be convenient to provide complementary data about human locomotion, such as the adaptation of stride length/stride frequency/walking speed to the incline of the terrain, stride-to-stride variability, and biomechanical efficiency.

The complete Global Positioning System (GPS) consists of 24 operational satellites and provides 24-h, all-weather navigation and surveying capability worldwide (15). The satellites transmit at two frequencies modulated with two types of codes: precise (reserved to U.S. army) and standard (free access). For both codes, a navigation message is encrypted, containing the satellite ephemeris, which allows the calculation of the position of the satellites. The position of the GPS receiver can be calculated from the pseudorange, which equals the distance between the satellite and the receiver deduced from the measured travel time of the code plus corrective terms. Given the geometric positions of the satellites (satellite ephemeris), four pseudoranges are sufficient to correct the clock error and to compute the 3D-position of the receiver with an accuracy of about 10 m. The accuracy obtained by stand-alone code mode is not sufficient for a precise analysis of locomotion (21). The method can be improved using a differential mode implying the use of one fixed GPS reference station and one moving receiver, with a precision of a few meters. In a previous study, we have validated the speed measurement accuracy by differential GPS (DGPS) in code mode (22). The error obtained on average walking and running speeds was within 0.1 km·h^{-1}. By processing short baselines with the carrier phase (GPS second observable) in differential mode, it is possible to eliminate most of the errors and to lead to a centimeter accuracy. The principle of the method is that a phase shift

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exists between the carrier wave of the GPS satellite and a reference wave generated by the GPS receiver.

Recent technological developments permit one to use GPS at a high sampling rate (5 to 20 computed positions each second) in differential phase mode. With this method, the accuracy is so high that it is not possible to validate instantaneous speed measurements under field conditions with classical distance and time measurement. Namely, because of the high correlation between two consecutive samples, it can be assumed that the speed is assessed with a theoretical precision of 0.07 $km \cdot h^{-1}$ every 0.2 s (19). We recently utilized GPS technology to assess trajectory, slope of the terrain, and walking speed on an outdoor circuit. It was possible to record every second the slope and the speed of walking in totally free subjects (18). We concluded that GPS receivers in differential phase mode seemed convenient for tracking unconstrained walking activities in humans. In addition, we tested in another study whether 5-Hz position measurements permitted us to retrieve the stride frequency (SF) by making a comparative assessment of SF by means of an accelerometer. A correlation of 0.998 ($P < 0.001$) was observed between GPS and accelerometer SF measurements (24).

Because it seemed that the temporal pattern of walking was correctly assessed by GPS, we decided, as a subsequent challenge, to attempt to measure the external mechanical power of walking. One should remember that the mechanical energy expenditure of the body during walking can be explained by the internal work (muscular energy spent to move the body's segments compared with the center of mass) and external work (energy spent to move the body's center of mass). With a GPS receiver and an antenna attached to the trunk, only the external work (potential and kinetic energy variations) can be assessed. The parameters we chose to calculate from GPS positions were the potential energy ($PE = mgh$), the kinetic energy ($KE = 0.5\ mv^2$), and the total energy ($E = PE + KE$). We also calculated the average vertical lift of the trunk per step and the average speed variation per step. From the energy data, the average power (potential, kinetic, and total) dissipated during the exercise was computed. The energy exchange (recovery) between potential and kinetic energy was also calculated. In addition, we simultaneously measured the metabolic energy expenditure by indirect calorimetry in order to calculate the mechanical efficiency of walking.

The results were compared with values obtained by other authors in the past under laboratory conditions. We chose to collect clues to demonstrate that the GPS receiver can provide positioning data with sufficient accuracy to be used in mechanical power analyses of outdoor walking. We hope that those preliminary results will encourage other research groups to truly validate the GPS technique as a new tool for obtaining information on biomechanical parameters.

METHODS

Subjects. The subjects were five individuals (three men and two women) whose average characteristics were as follows (mean ± SD): age, 26.0 ± 2.1 yr; body mass, 65.4 ± 9.3 kg; body height, 173.6 ± 10.6 cm; and body mass index (BMI), 21.6 ± 0.9 $kg \cdot m^{-2}$. They were informed about the course of the experiment and signed an informed consent form according to the institutional guidelines and the Declaration of Helsinki.

Procedure. Before and after the experiment, each subject lied down for 20 min wearing a portable calorimeter for assessing the resting metabolic rate (RMR) (23). The average RMR was 73.2 ± 18.1 W.

Each subject walked five times for 5 min on a standard outdoor athletic track at different stride frequencies. Between walks, a 2-min break was imposed in order to allow the subject to return close to a RMR. The subjects were asked to regulate their walking pace by following the beep of an electronic metronome. Selected frequencies were 70, 90, 110, and 130 $steps \cdot min^{-1}$ (i.e., 1.17, 1.5, 1.83, and 2.17 Hz). This method allowed a wide range of walking speeds (2.8 to 6.30 $km \cdot h^{-1}$) while keeping a constant gait style and a steady state energy expenditure (EE). The fifth run was walked at a self-selected pace: the instructions were to choose the most comfortable stride rhythm and to keep a constant gait style.

Instrumentation. The precise positioning of subjects was realized with two Leica System 500 double frequency GPS receivers (Leica Geosystems, Heerbrug, Switzerland), measuring at a 5-Hz rate. The GPS logger (rover) was placed in a backpack tightly fixed to the person. The receiving antenna, placed just over the head of the subject, was mounted on a metal bar fixed onto the dorsal pack. Differential carrier phase localization between the fixed base station and the antenna mounted on the walking person was used. The results were computed in postprocessing mode using Leica's SKI-Pro software. The total weight of the GPS equipment was about 4 kg. For measuring the oxygen consumption and the production of carbon dioxide, a fully portable breath-by-breath pulmonary gas exchange system was used ($K4b^2$, Cosmed, Italy (13)). This apparatus included a digital bidirectional turbine (embedded in a face mask) for the measurement of the pulmonary ventilation. A thermomagnetic O_2 analyzer and an infrared CO_2 analyzer measured the gas concentration in the expired air. The total weight of the equipment was about 0.8 kg. The subsequent calculations were made by using the Matlab program (MathWorks, Inc., Natick, MA).

Data analysis. EE was calculated from $\dot{V}O_2$ and $\dot{V}CO_2$ using standard equations of indirect calorimetry (23). Breath-by-breath data were averaged for the 2 last min of each test, which were considered steady-state periods. RMR was calculated by averaging 5 min of each of the two resting periods before and after the experiment. EE induced by activity (net EE, EE_{net}) was calculated by subtracting RMR from gross EE (EE_{gross}).

GPS data were synchronized with calorimeter data: the last 2 min of each exercise was therefore analyzed. The speed of the GPS antenna in each 3D direction was computed from 3D position data by numeric derivation.

The vector magnitude of the speed was then calculated for each 5-Hz sample. We assumed that the displacement of the GPS antenna reflected the trunk movements, which can be considered similar to the movements of the center of mass (see Discussion, below). The average vertical lift of the trunk per step (Δz) and the average speed variation per step (Δv) were calculated by accumulating throughout the positive vertical displacements and the positive speeds:

$$\Delta z = \frac{\sum((z_{t+1} - z_t) > 0)}{T \cdot SF} \qquad (1)$$

$$\Delta v = \frac{\sum((v_{t+1} - v_t) > 0)}{T \cdot SF} \qquad (2)$$

where z_t is the vertical GPS position at time t, v_t the speed, T the total time (120 s), and SF the stride frequency (estimated by Fourier transform of the vertical signal). The potential (PE) and kinetic (KE) energy components of the body at time t were calculated and summed to yield a total mechanical energy value:

$$E_t = PE_t + KE_t = (mgh)_t + 1/2(mv^2)_t \qquad (3)$$

where m is the body mass, g is the acceleration induced by gravity, h is the vertical position of the antenna above the lowest point in the whole 2-min record, and v is the resultant (vector magnitude) of the 3D speeds.

We hypothesized that the variation of the mechanical energy approximately followed a sinus wave at step frequency, and therefore that total frequency content was within the 0- to 2.5-Hz range (see Discussion, below). Consequently, in order to obtain a better temporal definition, we interpolated the energy signals (PE, KE, and E) by using the FFT method. Hence, we obtained signals at five times the original sampling rate (25 Hz). The average mechanical power of each exercise was calculated by considering only the positive variation in energy (namely, positive work):

$$W_k = \frac{\sum((KE_{t+1} - KE_t) > 0)}{T} \qquad (4)$$

$$W_p = \frac{\sum((PE_{t+1} - PE_t) > 0)}{T} \qquad (5)$$

$$W_e = \frac{\sum((E_{t+1} - E_t) > 0)}{T} \qquad (6)$$

where W_k, W_p, and W_e are, respectively, the average kinetic, potential, and total power in watts, PE, KE, and E the energy as calculated in equation 3, and T the total time (120 s). We also assessed the energy recovery R induced by the time-shift between potential (W_p) and kinetic (W_k) energy, as defined by Cavagna and Kaneko (6):

$$R = \frac{W_k + W_p - W_e}{W_k + W_e} \cdot 100 \qquad (7)$$

RESULTS

Figure 1 presents a typical result of the calculation of the mechanical energy. Raw 5-Hz data are presented superimposed to the interpolated 25-Hz signals. In order to better illustrate the GPS capabilities, we chose to display a portion of the signal corresponding to the end of a walking exercise. A clear temporal shift exists between kinetic and potential energy: high peaks in one signal correspond to low peaks in the second one. This shift is the illustration of the well-known mechanism by which the energy is spared during walking (energy exchange or recovery). When the subjects stopped walking, the kinetic energy decreased to zero and the potential energy stabilized to a value corresponding to the "standing still" altitude. It should be noted that the zero altitude level was arbitrarily taken as the lowest altitude in the whole sample, which is not in the portion of the data presented here.

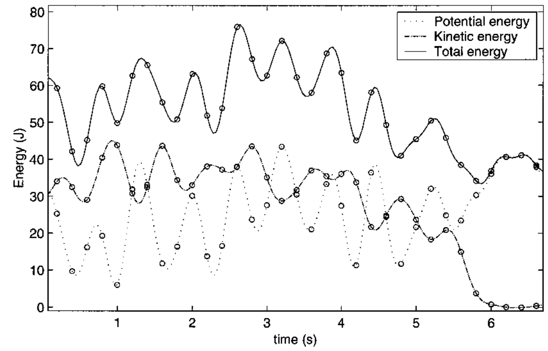

FIGURE 1—External mechanical energy during walking. One subject (woman, 23 yr, 160 cm, 54 kg) walked at 1.03 m·s^{-1} (90 steps·min^{-1}). For a better presentation, only 7 s of a 5-min record is shown. The subject stopped walking at 5.5 s. Potential (mgh), kinetic (0.5 mv^2), and total energy were computed for every 5-Hz GPS sample. *Open circles* are 5-Hz data. *Dashed*, *dotted*, and *continuous lines* represent 25-Hz interpolated signals.

Table 1 summarizes the averaged results ($N = 5$ subjects) in function of imposed stride frequencies. The mechanical power and the energy expenditure increased with the walking speed: we found a good linear relationship between EE_{gross} and total mechanical power ($r = 0.95$, $P < 0.01$, $N = 25$, i.e., five subjects and five different walking paces). These results indicate that mechanical power assessed by GPS follows, as expected, a parallel increase with the metabolic energy expenditure.

Figure 2A displays individual results for the vertical lift of the trunk, which varied from 23 to 89 mm. The lift increased with speed (correlation coefficient between the average lifts and speeds: 0.97 ($N = 5$); $P < 0.05$). A clear relationship between body height and vertical lift was also found; for instance, at freely selected pace (1.56 m·s^{-1}), the correlation was 0.96 ($P < 0.05$, $N = 5$).

The energy exchange (recovery), described as the "rolling egg" or inverted pendulum model (6), seemed to decrease with walking speed. It exhibited a substantial interindividual variation, with a total range from 9 to 56% (Fig. 2B). Two subjects (mp, mg) had relatively constant recovery for each speed. In contrast, subjects vg and ql exhibited a substantial intraindividual variation of the recovery: the range was 9 to 33% and 19 to 56%, respectively.

TABLE 1. Mechanical power and energy expenditure in walking.[a]

Imposed Stride Frequency (steps·min^{-1})	70	90	110	130	Free
Walking speed (m·s^{-1})	0.70 ± 0.09	1.05 ± 0.11	1.45 ± 0.18	1.72 ± 0.22	1.56 ± 0.17
Gross energy expenditure (W)	168.0 ± 41.2	216.7 ± 55.1	295.7 ± 93.4	381.6 ± 120.5	324.1 ± 76.4
Net energy expenditure (W)	95.0 ± 27.0	143.7 ± 40.2	222.7 ± 79.2	308.6 ± 107.5	251.1 ± 62.4
Vertical lift per step (mm)	34 ± 9	43 ± 14	54 ± 17	67 ± 15	67 ± 16
Potential power (W)	27.1 ± 9.7	46.0 ± 20.4	71.0 ± 30.1	97.0 ± 36.2	89.8 ± 33.8
Speed variation per step (m·s^{-1})	0.21 ± 0.06	0.15 ± 0.03	0.12 ± 0.01	0.13 ± 0.02	0.13 ± 0.02
Kinetic power (W)	12.2 ± 6.8	16.9 ± 7.2	23.1 ± 6.6	32.2 ± 6.8	28.0 ± 10.1
Total mechanical power (W)	22.8 ± 7.9	38.9 ± 18.3	70.2 ± 34.1	99.5 ± 46.2	83.5 ± 39.7
Recovery (%)	40.5% ± 11.5	39.1% ± 10.0	28.1% ± 11.4	25.7% ± 13.1	31.3% ± 11.8
Gross mechanical efficiency (%)	13.6% ± 4.1	17.6% ± 5.8	23.0% ± 7.5	25.4% ± 6.3	24.6% ± 6.5
Net mechanical efficiency (%)	24.5% ± 8.5	26.9% ± 9.8	30.8% ± 10.0	31.8% ± 8.1	31.8% ± 8.0

[a] Five subjects walked four different runs at increasing stride frequency, and one ran a freely selected pace. Values are means ± SD. Walking speed and mechanical energy were calculated from GPS recording as explained for Figure 1. Mechanical power was calculated by integrating the positive work and dividing the result by the time. Vertical lift of the trunk per step was calculated from cumulated vertical positive displacement, total time, and step frequency. Speed variation per step was computed from cumulated positive speed variation, total time, and step frequency. Recovery is the energy exchange between potential and kinetic energy. EE_{gross} (gross energy expenditure) was computed from indirect calorimeter data. EE_{net} is EE_{gross} minus resting metabolic rate. Gross work efficiency is mechanical power divided by EE_{gross}. Net work efficiency is mechanical power divided by EE_{net}.

DISCUSSION

The present study represents a challenging application of satellite positioning technique. The purpose was to illustrate the potential use of a high-precision GPS receiver for locomotion analyses. In previous studies, we found that GPS, in differential phase mode, can track subjects to obtain different parameters such as position, speed (21,22), the incline of the terrain (18), and stride length and frequency (24). The subsequent step was to attempt to collect more parameters from position data. Energy information, such as external mechanical power, is directly derived from displacement data, such as vertical lift of the trunk. It was therefore logical to go ahead into this research field.

As pointed out by Winter (25), the measurement of the displacement of the trunk segment does not constitute the optimal solution to study the kinematics and biomechanics of the locomotion, because the exact movement of the center of gravity is not assessed by this method. However, recent studies have demonstrated the usefulness of the assessment of the vertical lift of the trunk to evaluate the biomechanical performance (11,14,20). Kerrigan et al. (14) concluded in their article: ". . . there is a role for bioengineering researchers to design a more accessible tool to measure the rise and fall of the trunk (or pelvis) during gait. With some innovative engineering, a nongait laboratory or bedside analysis could become possible." No mention was made of GPS applications at that time. Saini et al. (20) have recently shown that the vertical movement of the center of mass, approximated by the sacral displacement method (vertical displacement of the trunk), was comparable to the assessed movement obtained by the reconstructed pelvis method or the segmented analysis method (determination of the real trajectory of the center of mass). It should be added that the classical methods for assessing the center of mass trajectory during locomotion are derived from several assumptions or exhibit inherent imprecision. For instance, segmented analysis is derived from a model, which fails to entirely fit with the reality (20). For outdoor free-living experiments about human locomotion, we believe that the apparent lack of accuracy of the GPS method proposed in this article is acceptable as compared with the advantage to be close to a physiological situation. Moreover, the GPS approach allows one to record and easily analyze an almost unlimited number of gait cycles (about 3–4 h of recording, limited only by memory card and battery autonomy). This is very difficult to realize with other techniques, except maybe with the "cinematic arm" developed by Belli et al. (3) for treadmill experiments.

During walking, the trunk goes up and down rhythmically. Therefore, the vertical displacement follows a periodical pattern whose waveform was described as a sinusoid at step frequency (8,14,20). Because the potential energy is the position multiplied by a constant term (mass and gravity acceleration), the same waveform is expected, such as that measured by Winter (25). Cavagna and Kaneko (6) and Cavagna and Margaria (7) described a comparable waveform for work against inertia. In order to consolidate the hypothesis that the trunk variation position could be approximated by a sinusoid at step frequency, and therefore that low sampling frequency (5 Hz) did not generate substantial errors, a pilot experiment was designed with a new GPS receiver recording positions at 10 Hz (Javad Positioning System, San Jose, CA). One subject (29 yr, 177 cm, 89 kg) walked at a freely selected speed for 30 s on an athletic track. In order to study the frequency content of the GPS signal, the power spectrum density of the 10-Hz vertical position (altitude) and walking speed (calculated by numeric derivation from horizontal positions) was computed by using Welch's averaged periodogram method (Matlab command *PSD*). The results are shown in Figure 3. The large majority of the signal power was found at step frequency. The subsequent harmonics were very faint. This confirms that the vertical displacement of the trunk or the speed variation are close to a sinus wave at step frequency. Despite the low sampling rate of position measurements (5 Hz), it was therefore possible to reconstruct the signal to the extent to which the half-sampling rate limit (2.5 Hz) was respected (Fig. 1). However, it should be recognized that the combination of position errors and low sampling rate could lead to a false estimate of the energy signal: the positive power may be over- or underestimated from one step to another. If these errors are assumed to have a gaussian distribution, the

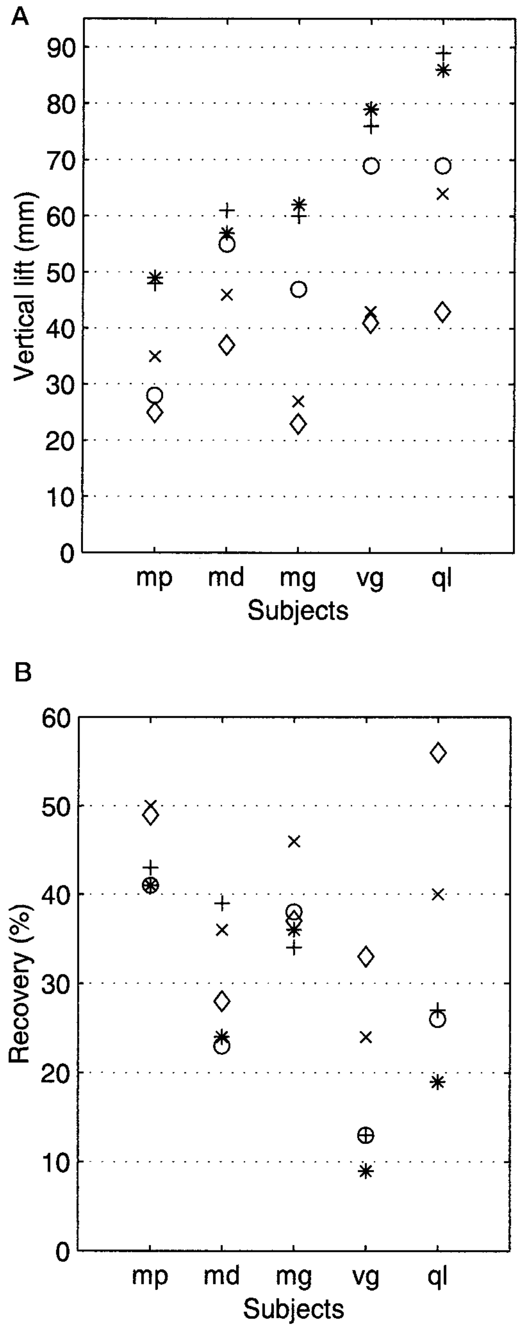

FIGURE 2—Individual results of the vertical lift of the trunk per walking step (A) and of recovery (energy exchange) between potential and kinetic energy (B). For each walking exercise and each subject, cumulated positive vertical displacement divided by total time and stride frequency was calculated (A). The energy exchange between the two energy components was calculated as explained in equation 7 (B). Subjects were sorted out by body height: mp, 160 cm; md, 167 cm; mg, 174 cm; vg, 180 cm; ql, 187 cm. $^{+}$free; *130 steps·min^{-1}; ^{o}110 steps·min^{-1}; ^{x}90 steps·min^{-1}; ◇70 steps·min^{-1}.

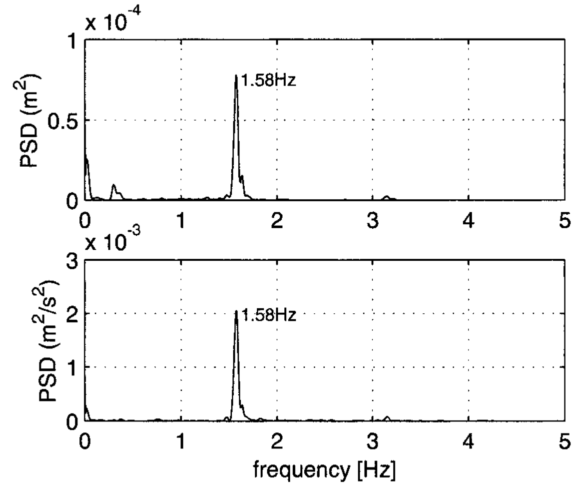

FIGURE 3—Power spectrum density (PSD) of the altitude and speed GPS signals. One subject (man, 29 yr, 177 cm, 89 kg) freely walked for 30 s at 4.5 km·h^{-1}. Altitude was directly derived from GPS 10-Hz data. Speed was obtained by numeric derivation of horizontal positions.

impact on the average power values is reduced by the fact that a substantial number of strides were analyzed (about 250 steps).

When performing walking studies under free-living conditions, a different protocol than that used for treadmill experiments has to be designed. In order to impose a constant rate of energy expenditure, we chose to keep the stride frequency constant, and let the subject spontaneously select the stride length and hence his or her walking speed. We are aware that this method is not totally comparable with the classical one, for which the walking speed is fixed and for which the subject self-adjusts the stride length/frequency. Consequently, one should keep this fact in mind when interpreting our results and comparing them with previous findings obtained under laboratory conditions.

The range of the vertical lift of the trunk per step we found (23–89 mm; Fig. 2) was slightly higher than that published by Kerrigan et al. (14) (10–60 mm; speed range, 0.45–1.79 m·s^{-1}), Saini et al. (20) (15–45 mm, free speed), Cavagna et al. (8) (10–55 mm; speed range, 0.42–2.5 m·s^{-1}), or Bhambhani et al. (4) (32 mm; speed, 1.33 m·s^{-1}). The possible overestimation of the lift could be explained by the fact that the GPS receiver and the antenna were not specifically designed for human locomotion analysis. Indeed, in order to explore the potential utilization of GPS before developing an adapted instrument, we used a commercially available professional GPS receiver without any ergonomic transformations. Such a device mounted on a backpack is commonly used for high-precision positioning for topographic measurements made by foot. The main drawback of this apparatus was that the antenna, which must be over the head of the subject for the satellite access, was not totally interdependent with the body: the system design did not allow the dorsal pack and antenna to be totally rigidly attached to the trunk. The inertia of the rucksack and the long metal bar to sustain the antenna may have added an extra oscillation, whose amplitude was most likely added to trunk movements. Other explanations about the overestimation of the vertical lift could stem from the slightly bouncing surface of the athletic track, which can produce a different gait style than treadmill walking. In addition, we can assume that the errors in GPS positions are not the main source of error, considering the excellent determination of the temporal pattern (24), the shape of the energy signals (Fig. 1), and the large size of the gait analysis sample that decreases the impact of errors on the average value. An additional indication about the GPS accuracy was that a good significant correlation between the vertical lift and the body height was observed ($r = 0.96$), such as explained by the compass gait model (14).

The time-shift between kinetic and potential energy variations is the main mechanism by which the energy output of the body is spared during walking. The GPS technique was able to highlight this well-known phenomenon. However, the average magnitude of the recovery was lower (20–40%) than described by Cavagna and Kaneko (6) (60–80%) or Minetti et al. (17) (15–85%). We think that the extra oscillation of the antenna modified the time-shift pattern of the work signal and hence induced a low energy exchange. This hypothesis is reinforced by the observation that the highest recovery among the subjects was systematically noted at low speed, for which one could assume that the extra oscillation of the antenna is minimal. Because the recovery varied substantially among the subjects (10–60%), it is unlikely that GPS constituted the main source of the error, since a measurement error would induce a systematic bias, whereas we observed substantial variability among subjects. The perturbing movement of the antenna may be more intense for some subjects, depending on their particular gait style.

Past experiments (10,14), conducted on the basis of trunk displacement measurements, have shown a close relationship between external mechanical work against gravity and metabolic energy expenditure. The metabolic energy expenditure and mechanical power are known to exhibit a second-degree relationship in function of the walking speed (8,10). Our results, showing an excellent correlation between EE_{gross} and mechanical work ($r = 0.95$, $P < 0.01$), are in accordance with these previous findings.

Few numerical results are available in the literature about the average external mechanical power of normal gait. Moreover, various methods, calculation techniques, and units have been used. One important issue is how the positive energy variations (positive work) and the negative energy variations (negative work) are considered. Indeed, positive work and negative work do not induce the same metabolic energy demand. For the present study, we decided, as have some other authors (12,17), to take into account positive work only. Using triaxial acceleration measurement of the trunk, Gersten et al. (12) found an external power of 282 $kg \cdot m^{-1} \cdot min^{-1}$ at a walking speed of 1.4 $m \cdot s^{-1}$, namely, 0.57 $W \cdot kg^{-1}$. On the basis of video analysis and a center of mass kinematics approach, Martin et al. (16) reported a mechanical power (negative and positive work) of 0.85 $W \cdot kg^{-1}$ (0.43 $W \cdot kg^{-1}$ of positive power) at 1.7 $m \cdot s^{-1}$. With a comparable approach, Winter (25) presented an average total work of 98 J per stride (negative and positive work) measured at various walking speeds, which corresponds to 0.69 $W \cdot kg^{-1}$ of positive power. Minetti et al. (17) showed an external positive power of 0.409 $W \cdot kg^{-1}$ at 1.5 $m \cdot s^{-1}$. Our results, on the basis of the measurement of trunk displacement, showed an average power of 0.60 $W \cdot kg^{-1}$ at 1.05 $m \cdot s^{-1}$ and 1.09 $W \cdot kg^{-1}$ at 1.45 $m \cdot s^{-1}$. Because vertical lift values were close to the previously published results, the origin of the overestimation was probably the low recovery between the two energy components. For instance, a subject who had a relatively high recovery ratio (41%) exhibited a mechanical power of 0.49 $W \cdot kg^{-1}$ at 1.31 $m \cdot s^{-1}$. In contrast, another subject with a low recovery (13%) produced a mechanical power of 1.36 $W \cdot kg^{-1}$ at 1.58 $m \cdot s^{-1}$.

Various values of work efficiency of walking are found in the literature because of a large variety of experimental protocols, definitions of efficiency, and methods of analysis (2,7–8,26). We calculated the work efficiency from data of a recent article published by Minetti et al. (17), who measured external positive power and EE_{net} at different walking speeds on a treadmill: the results were an efficiency of about 12% between 0.75 and 1.72 $m \cdot s^{-1}$. Our results were higher (25–32% work efficiency), which is a direct consequence of the overestimated total mechanical power. It should be noted that the subject with the highest recovery ratio had a net mechanical efficiency ranging from 17–22%. In contrast, the subject with the lowest recovery exhibited extremely high efficiency values (27–38%).

Although a full validation procedure remains to be performed, we believe that new GPS technology is able now to reach a sufficient accuracy to record small 3D movements of the body during walking. Temporal pattern of each gait cycle can be retrieved, although current technical design of the GPS antenna (not yet adapted to biomechanics) probably led to an underestimation of the energy exchange between potential and kinetic energy. Consequently, the total mechanical power and the efficiency may have been overestimated as compared with the previously published values derived from other methodologies. The use of an antenna rigidly fixed to a helmet will probably partially solve this issue, but this remains to be further studied. New high-sampling-rate GPS (20 Hz) will permit the recording of more position points for each gait cycle in order to solve potential errors induced by low sampling rate. When an ergonomic high-frequency GPS receiver becomes available, a true validation study will be performed, for instance, by comparing GPS and a video motion analysis system. We are confident that GPS technology will open up new perspectives in the field of applied biomechanics by bringing the research laboratory outdoors, near a real-life situation.

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