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Validity of the Pedar Mobile system for vertical force measurement during a seven-hour period

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Abstract

Objective measurement of weight bearing during a long-term period can give insight into the postoperative loading of the lower extremity of orthopedic patients to avoid complications. This study investigated the validity of vertical ground reaction force measurements during a long-term period using the Pedar Mobile insole pressure system, by comparing it with a Kistler force platform. In addition, the validity of a new sensor drift correction algorithm to correct for offset drift in the Pedar signal was evaluated. Ground reaction force data were collected during dynamic and static conditions from five healthy subjects every hour for 7 h. A mean offset drift of 14.6% was found after 7 h. After applying the drift correction algorithm the Pedar system showed a high accuracy for the second peak in the ground reaction force-time curve (1.1 to 3.4% difference, $p>0.05$) and step duration (−2.0 to 4.4% difference, $p>0.05$). Less accuracy was found for the first peak in the ground reaction force–time curve (5.2 to 12.0% difference; $p<0.05$ for the first 3 h, $p>0.05$ for the last 4 h) and, consequently, in the vertical force impulse (5.5 to 11.0% difference, $p>0.05$). The Pedar Mobile system appeared to be a valid instrument to measure the vertical force during a long-term period when using the drift correction program described in this study.

Keywords: Pedar Mobile system; Validity; Vertical force; Drift correction; Long-term period

1. Introduction

To avoid complications, instruction on partial weight bearing is often given during the rehabilitation of orthopedic patients with various pathologies of the lower extremity (Chow and Cheng, 2000; Endicott et al., 1974; Gapsis et al., 1982; Huiskes, 1998; Perren and Matter, 1996; Phillips et al., 1991; Siebert, 1994; Tveit and Kärrholm, 2001; Weaver, 1975; Wirtz et al., 1998). It is evident that the ground reaction forces under the foot during weight bearing (i.e. when walking and standing) generate forces and moments in other structures in the lower extremity, such as the hip (Bergmann et al., 1993; Davy et al., 1988). In daily clinical practice, because the forces in the hip cannot be directly measured, the ground reaction force under the foot is used as a load measure, often expressed in percentage body weight. Patients are generally instructed to perform partial weight bearing during a period of 6–8 weeks. To evaluate the effectiveness of this instruction and to quantify the loading of the lower limb during the day, objective measurement is needed of the actual amount of loading (vertical ground reaction force) and other aspects of loading (i.e. step duration, vertical force impulse) during weight bearing, both in and outside the clinic, and during a long-term period.

Portable insole pressure devices can measure the actual amount of load bearing during daily activities and over a long-term period (hours) (Hurkmans et al.,

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2003). However, the validity of vertical force measurements performed by insole systems (especially during long-term periods) may be influenced by temperature or humidity in the shoe, and by loading of the sensors during an entire day (Cavanagh et al., 1992). Moreover, insole sensors measure the "normal" force, which is not necessarily similar to the vertical ground reaction force (Kalpen and Seitz, 1994; Kernozek et al., 1996; McPoil et al., 1995). Only a few studies have used an insole pressure system for long-term measurements (Abu-Faraj et al., 1997; Perren and Matter, 1996; Siebert, 1994; Tveit and Kärrholm, 2001). Perren and Matter concluded that their insole system (based on a hydraulic principle) was not technically reliable enough for routine use in the clinic. Discrete insole pressure systems, developed by Tveit and Kärrholm (2001) and Abu-Faraj et al. (1997), have some disadvantages compared to matrix insole devices: the transducer may act as a foreign body in the shoe, and inaccuracies may occur due to imprecise positioning of the sensors (Abu-Faraj et al., 1997; Cavanagh et al., 1992; Lord, 1981). No reports were found on the validity of these insole systems to measure the vertical ground reaction force during long-term measurements.

Arndt (2003) performed long-term measurements using Pedar pressure matrix insoles (Novel GmbH, Munich, Germany) and found a 17% sensor creep after 3 h. To correct for this creep, Arndt presented a correction method in which short standing trials were used to reset the signal based on the assumption that the measured body weight does not change during the trial. In that study, no data were presented regarding the validity of the Pedar system after the correction method was used. However, for long-term partial weight bearing measurements, we believe that Arndt's correction method is not optimal because the patient uses a walker or crutches meaning that the total body weight cannot be measured. In the present study, we introduce a drift correction algorithm to correct for the possible offset drift during walking in the Pedar mobile system.

The aim of this study was to investigate the validity of the Pedar Mobile system to measure vertical force during a long-term period. The main research questions were: What is the amount and type of drift when using the Pedar system for 7 h? How accurate is the Pedar system in measuring vertical force over a long-term period when corrected for possible offset drift?

2. Methods

2.1. Subjects

Five healthy subjects (3 females and 2 males) with an age range of 21–35 years (mean 26 years) and weight range of 60–89 kg (mean 69 kg), participated in the study. None of the subjects had a history of musculoskeletal trauma or disease of foot or ankle. An overview of the subjects' characteristics is presented in Table 1.

Table 1
Subject characteristics at $t = 0$ h

Subject	Gender	Age (yrs)	Weight (kg)	Insole (type)	Shoe size (European)
1	Female	28	67	*W*	40
2	Female	22	59	*W*	40
3	Female	25	61	*W*	40
4	Male	35	70	*X*	42
5	Male	21	89	*X*	42

W = humidity-proof *W*-sized insoles; *X* = humidity-proof *X*-sized insoles.

2.2. Material

The Pedar Mobile system (a portable device with matrix insoles containing 99 capacitance sensors) was used to measure vertical force during a long-term period. Three custom-made battery units, consisting of two Sony NP750 Li-ion batteries, were needed to provide the Pedar system with power for an 8-h period. The Pedar start–stop trigger cable was used to record data during the measurement protocol after every hour. The in-shoe pressure data were stored on a 40 Mb PCMCIA flash card. Pedar mobile Expert version 8.2 software and a custom-written Matlab® correction program were used to analyze the data.

A Kistler (type 9281B12) force plate was used to measure the vertical ground reaction force. Data were collected from the platform via two Kistler type 5001 and two Kistler type 5011 charge amplifiers, and A/D conversion was done with a 12-bit resolution DASH-16 PC board. Custom-made data acquisition software was used to collect the Kistler data.

All subjects wore similar athletic running shoes: the men wore a shoe size 42 (European; UK: 7.5–8; US: 8.5–9) and the women a shoe size 40 (European; UK: 6.5; US: 8.5). For the purpose of this study, Novel GmbH developed humidity-proof versions of the Pedar insoles to decrease sensor drift during long-term measurements (these insoles are now commercially available). The men used humidity-proof X-sized insoles (shoe size 42/43) and the women used humidity-proof W-sized Pedar insoles (shoe size 40/41).

2.3. Protocol

The Pedar insoles were calibrated using the Trublu calibration device (Novel GmbH) and a GDH 14AN digital manometer (Greisinger Electronic GmbH, Regenstauf, Germany). The pressure loads applied were 4, 7, and 10 up to 60 N/cm^2 with intervals of 5 N/cm^2. Pedar data were collected for the right foot only using a sampling frequency of 99 Hz. The Kistler force data were recorded using a sampling frequency of 500 Hz. Before measurement, the Pedar Mobile system was turned on 1 h in advance (acclimatization period) and zero settings were done at $t = 0$ and $t = 1$ h. In preliminary tests we found a negative drift in the Pedar system data, which stabilized after 1 h; based on this, the Novel company recommended an acclimatization period of 1 h after which a second zero setting should be performed. After the second zero setting at $t = 1$ h, dynamic and static measurements were performed every hour for 7 h. For the dynamic measurements each subject walked at their own walking speed and positioned themself in front of the force plate so that the third right footstep was placed on the platform (Wearing et al., 1999; Miller and Verstraete, 1996). This was repeated 10 times for each subject every hour. For the static measurements, the subjects stood still on the left leg only, followed by standing still on the platform for 10 s on the right leg only. To standardize the subject's activities during the 7-h period, most of the time the subjects were sitting behind a computer, but during each hour were asked to stand up at least 5 times and on two occasions each hour to walk about 10 m.

2.4. Data analysis

Pedar mobile Expert version 8.2 software was used to calculate the force data from the Pedar system. Then, all Pedar and Kistler data were imported in Matlab® and were filtered using a low-pass Butterworth filter with a cut-off frequency of 40 Hz. In the analysis, drift was defined as an undesired change in output signal (force) over a period of time that is unrelated to the input (load). The drift was expressed as (1) absolute drift, defined as the increase in force measured during the unloading periods of the insoles (i.e. during swing phase for the walking trials and during the time the right leg was in the air for the standing trials), and (2) relative drift, defined as the increase in force during unloading periods and expressed as a percentage of the force during loading periods (static and dynamic measurements). Next, the type of drift (offset or gain) was assessed, because our correction method assumes that the drift is an offset drift. Offset drift was defined as a drift in which all output values (during loading and unloading periods) are increased at a certain time by the same value. This offset drift is relatively easy to correct, in contrast to gain drift in which output values are increased by a multiplication factor. To determine the type of drift, the drift measured with the right leg in the air (insole unloaded) was subtracted from the drift measured with the right leg on the force plate (insole loaded). If this difference (insole loaded minus insole unloaded) was constant over time, then the drift would be an offset drift and not a gain drift.

Offset drift during walking was corrected using a custom-made drift correction algorithm. The main steps of the automated correction algorithm are shown in Fig. 1. First, a threshold force value (which has to be above the maximum offset drift; Fig. 1A) was set to detect the first data point of the descending force curve of each step (referred to as "cycle detection point") below the threshold force value. This was done to define a unique point in each gait cycle. Secondly, the minimum force during each cycle was determined as the lowest force value between two consecutive cycle detection points (Fig. 1B). After this, a first-order polynomial was fitted between two consecutive minima by determining the x-value (time) and y-value (force) of

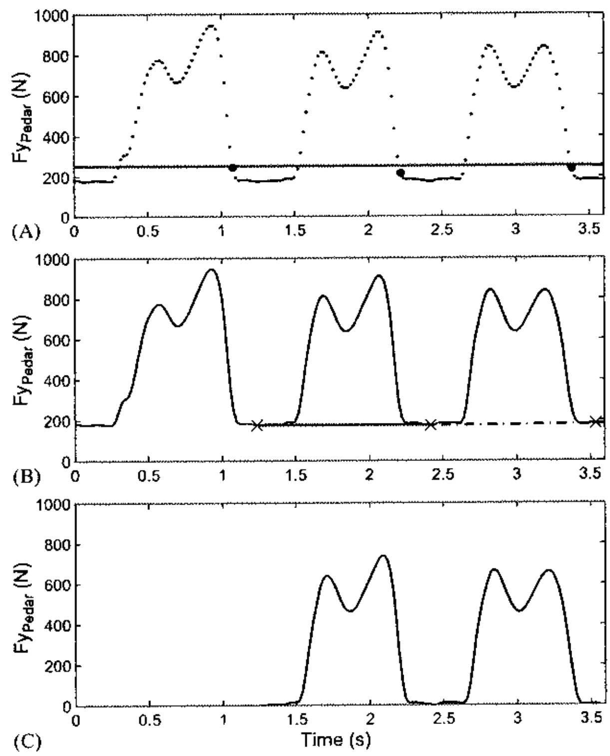

Fig. 1. Graphical representation of the algorithm used to correct for offset drift. As an example the Pedar force data of the first three right footsteps of subject 2 recorded at hour 7 are shown. See Methods (Data analysis) for a detailed explanation. (A) The threshold value (horizontal line) was set at 250 N, which was just above the maximum offset drift of approximately 200 N. The cycle detection point, indicated with ● and defined as the first sample of the descending force curve below the threshold, is shown for the three steps. (B) The minimum of each step, indicated with ×, was detected as the lowest force between two consecutive cycle detection points. A first order polynomial was fitted between two consecutive force minima (–·–·–) for step two and step three. (C) For each x-value (time), the y-value (force) of the polynomial was subtracted from the corresponding raw force data point to get the offset drift-corrected vertical force.

the first (x_1, y_1) and second minima (x_2, y_2), after which the slope coefficient (s) of the first-order polynomial was calculated using the following equation: $s = (y_2 - y_1)/(x_2 - x_1)$. The first-order polynomial equation was then $y = a + sx$, in which a was the y-value of the first minima point (y_1). Then for each x-value, the y-value of the polynomial was calculated and subtracted from the corresponding raw force data point to get the offset drift corrected vertical force (Fig. 1C). The correction method was based on the following assumptions: (1) the drift is an offset drift, (2) the drift between two subsequent steps is linear; and (3) the force during the swing phase is zero.

The accuracy of the Pedar Mobile system measurements after correction for offset drift was determined by the absolute and relative error of measurement. The absolute error was calculated as the difference between Kistler output data and Pedar Mobile output data; the relative error expressed the absolute error as a percentage of the force measured with the Kistler platform. Four variables that are important for weight bearing measurements were compared between the Kistler and Pedar system: (1) the first peak force in the M-shaped ground reaction force–time curve (N); (2) the second peak force in the ground reaction force-time curve (N); (3) the vertical force impulse (area under the force–time curve, N s); (4) the step duration (s). The mean and standard deviation were calculated for each of the paired data from the Pedar system and Kistler force plate. Paired t-tests were done using SPSS 10.1.0 for Windows. The level of significance for all tests was set at 5%.

3. Results

3.1. Amount and type of drift

The amount of drift found over 7 h for the dynamic and static measurements is presented in Table 2. The data generally showed minor drift for the first 3 h and an increase in drift after hour 4. The individual drift data for the dynamic measurements showed a relatively small drift for the first 4 h for subjects 1, 2, and 3, while drift increased from hour 4 to hour 7 (Fig. 2); these latter subjects were the three females with insoles W. The two male subjects (4 and 5), wearing insoles X, showed a larger drift which started at the beginning of the measurements. The mean drift after 7 h was 132 (14%) and 141 N (16%) for the walking and standing experiments, respectively.

The mean force curves for standing on the right leg (insole loaded) and for the right leg in the air (insole unloaded) showed a similar drift over 7 h (Fig. 3). At hour 1 the difference between the insole loaded and the insole unloaded was 728 N. At hour 4 and hour 7 the differences were 757 and 758 N, respectively; these differences were not significantly different from hour 1 ($p = 0.630$ and $p = 0.203$). Because there was no significant change in the difference between the measured force during loading and unloading of the insole between the measurements, this indicates that the drift was predominantly an offset drift.

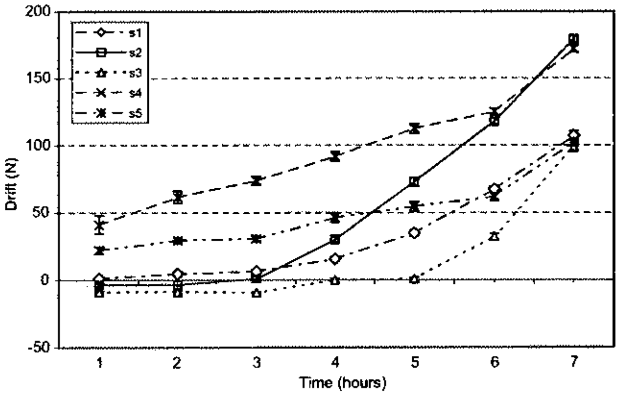

Fig. 2. The drift of Pedar force data for the five subjects (s1 to s5) over 7h, measured by the Pedar Mobile system during the swing phase of the dynamic measurements.

Table 2
Mean (standard deviation) of the absolute drift, and the relative drift found over 7h of the dynamic and static measurements with the Pedar Mobile system

	Hour 1	Hour 2	Hour 3	Hour 4	Hour 5	Hour 6	Hour 7
Walking							
Absolute drift[1] (N)	10.41 (20.77)	16.68 (29.03)	20.53 (33.22)	36.71 (35.27)	55.07 (41.68)	80.84 (39.41)	132.04 (40.03)
Relative drift, Fp1 (%)	1.39	2.21	2.68	4.48	6.59	9.33	14.17
Relative drift, Fp2 (%)	1.27	2.06	2.55	4.39	6.48	9.25	13.92
Standing							
Absolute drift[2] (N)	17.74(31.41)	21.29 (30.39)	30.69 (33.24)	51.01 (43.10)	59.08 (41.85)	100.08 (35.04)	141.57 (35.14)
Relative drift (%)	2.38	2.85	3.99	6.31	7.28	11.60	15.60

[1]Force measured during swing phase;
[2]Force measured during right insole unloaded; Fp1 = first peak force; Fp2 = second peak force

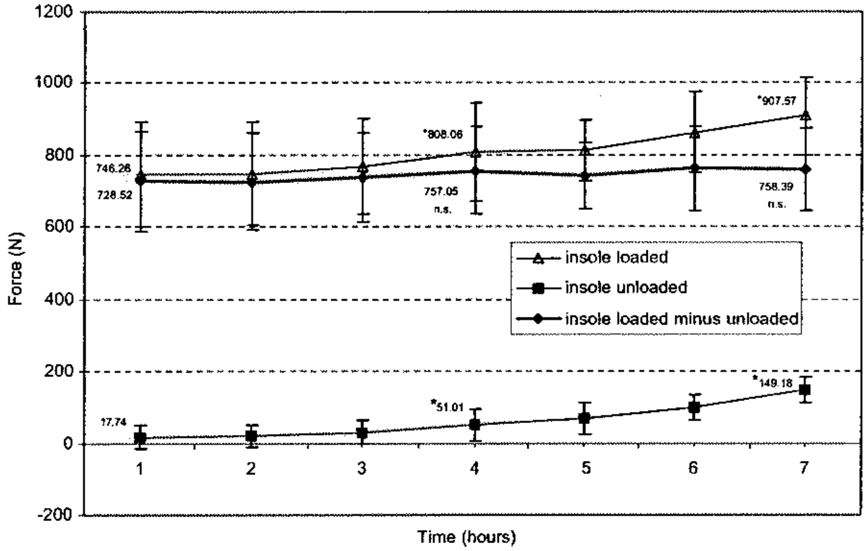

Fig. 3. The mean force and standard deviation measured by the Pedar Mobile system during standing on the right leg (insole loaded), during right leg in the air (insole unloaded) for the five subjects over 7 h. The difference between the "insole loaded" and the "insole unloaded" curve is presented by the "insole loaded minus insole unloaded" curve. * = significant difference from hour 1; n.s. = non-significant difference from hour 1.

3.2. Accuracy of the Pedar Mobile system with offset drift correction

After offset drift correction using the drift correction algorithm, the differences between Kistler and Pedar Mobile data were relatively small for the second peak in the ground reaction force–time curve. The relative errors ranged from 1.1 to 3.4%, and were not significantly different for all 7 h (Table 3; Fig. 4). The first peak force showed larger differences, with relative errors ranging from 5 to 12%. However, only for hours 1, 2, and 3 were these differences significant. All Pedar Mobile force data were lower than the Kistler force data. The vertical force impulse data as well as the calculated step duration were not significantly different from the Kistler data.

4. Discussion

This study investigated the amount and type of drift when the Pedar Mobile system was active for 7 h, as well as the validity of the Pedar Mobile system to measure vertical force over a long-term period when corrected for offset drift.

The Pedar data showed a drift of up to 14% when the system was active for a period of 7 h. During the dynamic measurements, the pair of Pedar insoles used by the three female subjects generally showed less drift for the first 3 h than the pair of insoles used by the two male subjects, while after 3 h the drift increased similarly in both groups (Fig. 2). These differences between insoles might arise because the insoles used by the females were new, whereas those of the males had been used for more than 6 months (Cavanagh et al., 1992). Hsiao et al. (2002) found that the accuracy and precision of a one-year-old insole were inferior to that of a new pair. In addition, in the present study the small differences between the male and female shoe types and/or the greater body weight of the male subjects (Arndt, 2003) may explain the different drift values. The somewhat larger drift in the static compared to the dynamic experiments (Table 2) might be explained by the measurement protocol, in which the dynamic trials were performed directly after the subjects had mainly been sitting with minimal sensor loading. The duration of sensor loading could also have influenced the measurements during standing (Hsiao et al., 2002 and McPoil et al., 1995). In the present study there was a relative sensor drift after 3 h of 3–4%, which is smaller than the 8–17% reported by Arndt (2003). This might be mainly related to the fact that Arndt studied two subjects walking constantly for 3 h wearing military boots and carrying a heavy load (49% of body weight), while our subjects where predominantly sitting.

Systematic comparison of static loading and unloading throughout the 7-h measurement period showed that the amount of drift was the same in both conditions, and that the difference between the loaded and unloaded condition remained constant. Based on this, we concluded that the drift was predominantly an offset drift, which was essential for the validity of the drift correction algorithm.

The correction algorithm is limited to walking data. We chose this method because the risk of putting too

Table 3
Comparison of the mean first and second peak force, vertical force impulse, and step duration over seven hours, measured with a Kistler platform and the Pedar Mobile system. Pedar Mobile force data are corrected for drift using the correction algorithm

	Hour 1	Hour 2	Hour 3	Hour 4	Hour 5	Hour 6	Hour 7
1st peak force (N)							
Kistler (sd)	833.88 (178.94)	838.54 (185.77)	835.14 (183.98)	847.79 (178.33)	836.23 (172.89)	837.10 (178.44)	843.78 (170.49)
Pedar (sd)	739.42 (150.19)	737.85 (153.30)	744.19 (141.17)	782.02 (131.86)	780.79 (127.81)	785.37 (130.02)	799.70 (118.65)
% difference	11.3	12.0	10.9	7.8	6.6	6.2	5.2
Mean difference (95% CI)	94.46 (32.6–156.3)	100.69 (29.1–172.3)	90.95 (31.7–150.2)	65.76 (−3.3–134.9)	55.44 (−9.6–120.5)	51.7 (−15.8–119.3)	44.1 (−29.4–117.5)
p–value[a]	0.013*	0.017*	0.013*	0.057	0.077	0.101	0.171
2nd peak force (N)							
Kistler (sd)	822.48 (188.30)	814.77 (197.60)	812.47 (194.40)	816.10 (190.47)	807.12 (173.28)	805.20 (179.90)	825.71 (190.22)
Pedar (sd)	810.51 (125.33)	793.30 (128.18)	785.13 (120.04)	799.94 (107.19)	794.85 (105.08)	792.75 (108.54)	816.32 (108.54)
% difference	1.5	2.6	3.4	2.0	1.5	1.6	1.1
Mean difference (95% CI)	11.97 (−90.8–114.8)	21.47 (−81.4–124.4)	27.34 (−87.9–142.9)	16.16 (−108.5–140.9)	12.27 (−104.3–128.9)	12.45 (−119.8–144.7)	9.39 (−127.7–146.5)
p–value[a]	0.763	0.594	0.546	0.737	0.785	0.807	0.858
Vertical force impulse (N s)							
Kistler (sd)	405.78 (85.70)	390.85 (76.07)	397.28 (75.22)	399.57 (78.47)	393.51 (74.96)	391.90 (75.71)	389.64 (81.13)
Pedar (sd)	359.70 (57.69)	350.46 (81.02)	353.57 (82.53)	368.99 (78.19)	365.79 (76.44)	365.87 (74.99)	368.13 (81.64)
% difference	8.1	10.3	11.0	7.7	7.0	6.6	5.5
Mean difference (95% CI)	46.08 (−36.9–129.1)	40.39 (−17.0–97.8)	43.7 (−11.8–99.2)	30.6 (−35.0–96.1)	27.7 (−31.4–86.8)	26.0 (−38.5–90.6)	21.5 (−44.0–87.0)
p–value[a]	0.175	0.122	0.094	0.265	0.263	0.326	0.413
Step duration (s)							
Kistler (sd)	0.70 (0.03)	0.70 (0.03)	0.71 (0.05)	0.71 (0.04)	0.71 (0.04)	0.71 (0.04)	0.70 (0.03)
Pedar (sd)	0.67 (0.15)	0.70 (0.18)	0.70 (0.18)	0.72 (0.16)	0.72 (0.14)	0.71 (0.15)	0.71 (0.16)
% difference	4.4	0.1	2.5	−1.5	−2.4	−0.1	−2.0
Mean difference (95% CI)	0.04 (−0.16–0.24)	0.00 (−0.16–0.16)	0.02 (−0.13–0.17)	−0.01 (−0.16–0.14)	−0.02 (−0.17–0.14)	0.00 (−0.13–0.13)	−0.01 (−0.15-0.13)
p-value[a]	0.598	0.992	0.755	0.860	0.783	0.988	0.794

[a] 5% significance level;
* = significant

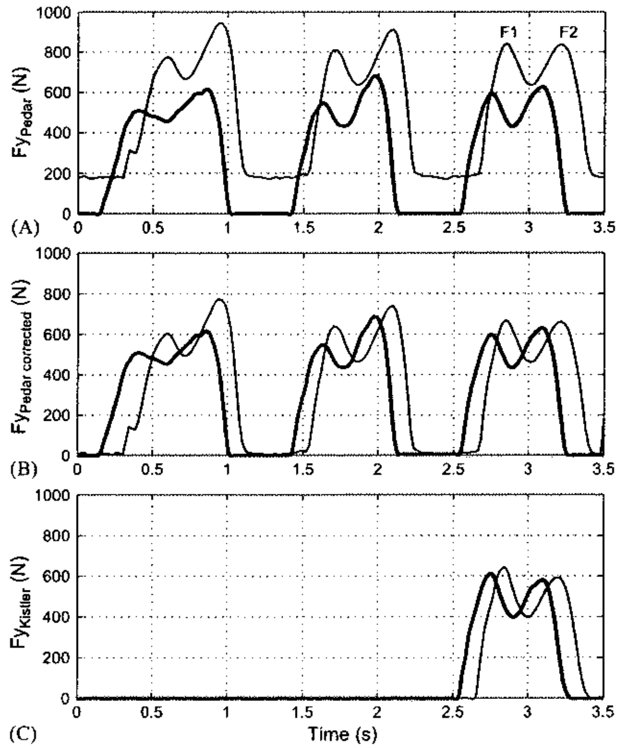

Fig. 4. Example of the offset drift correction of Pedar force data using the correction algorithm. (A) Pedar force data before offset drift correction of the first three right footsteps of subject 2, recorded at hour 1 (——) and hour 7 (—). (B) The same Pedar force data after offset drift correction. (C) Kistler force data of the third right footstep. F1 = first peak force; F2 = second peak force.

much weight on the lower extremity during partial weight bearing is probably much higher during walking than during standing. Although the measurements in this study were done using a capacitive insole pressure system, the drift correction method is independent of the measurement system.

The present study showed that a good estimate of the vertical force during walking can be obtained with the Pedar Mobile system during a long-term period after using the drift correction algorithm (Table 3). No significant differences were found between the Pedar and the Kistler data concerning the second peak in the ground reaction force, the vertical force impulse, and the step duration. The Pedar Mobile data systematically underestimated the first peak in the ground reaction force (5–11%). This underestimation is in line with studies by Barnett et al. (2000) and Boyd et al. (1997), who reported 14–16% lower Pedar values compared to force plate data. For the second peak in the ground reaction force, the accuracy found in the present study (relative errors of 1.1–3.4%) was higher than that reported by Barnett et al. (2000) (3–11%) and Boyd et al. (1997) (6%). The acclimatization period of 1 h, which we used to correct for negative drift (5–8%) before the measurements, might explain the higher accuracy in the present study. Although Barnett et al. (2000) mentioned an "acclimatization" in their study procedure, no specific information was given regarding duration or zero settings.

The underestimation of the first and second peak force in the Pedar Mobile system compared to the Kistler force data might be related to the way matrix sensors measure force compared to force platforms. The matrix sensors of the Pedar Mobile system measure the force perpendicular ("normal force") to each sensor (Kalpen and Seitz, 1994; Kernozek et al., 1996; McPoil et al., 1995). Therefore, especially during heel-strike and toe-off, the force vector of each sensor is different from the vertical force vector of the force platform. Generally, the sensors of the insoles are positioned more parallel to the force platform during toe-off compared to heel-strike, which might explain the higher accuracy of the second peak force measurements. The step duration measured by Pedar Mobile showed a high accuracy, comparable with data reported by Barnett et al. (2000). The vertical force impulse data were mainly influenced by the differences in the first peak force, because the differences in the second peak force and step duration were relatively small (Table 3).

Insole pressure measurements may be influenced by the type of footwear. For example, Barnett et al. (2000) found differences in vertical force data between running shoes (soft sole) and leather shoes (hard sole). However, other studies reported no differences in vertical force measurements between different shoe types (Bergmann et al., 1995; Clarke et al., 1983; Nigg et al., 1987; Nyska et al., 1995). In this study, the male subjects wore a different brand of shoes than the female subjects, but both brands were running shoes. The difference in shoe type would probably only have a major influence on the insole measurements if completely different types of shoes were used (Barnett et al., 2000). However, because the same type of shoe was used in this study their influence on the force measurements is considered as negligible.

Another aspect is that insole sensors measure the "normal" force, which is not necessarily similar to the vertical ground reaction force (Kalpen and Seitz, 1994; Kernozek et al., 1996; McPoil et al., 1995). It might be argued that the resultant force vector (F_r) is more comparable during heel-strike with the Pedar force vector than the vertical force (F_y). However, data presented by Winter (1991) show that during walking the mean F_y is 100% of body weight, and the mean F_x is 15–20% of body weight. Therefore, the calculated F_r would be 101–102% of body weight and thus does not differ much from the vertical force.

The lower force values measured with the Pedar system compared to the Kistler force plate could be due to the lower sampling rate, which was 99 Hz for the Pedar system and 500 Hz for the Kistler force plate. Visual inspection of the ground reaction force–time curves (see Fig. 4) indicated that, despite this difference, the force–time curves of both systems showed similar shapes for the first and second peak force, indicating that the lower sampling frequency of the Pedar system did not influence the measured peak forces.

The present study indicates that the Pedar Mobile system can be used in both clinical and research settings to evaluate vertical ground reaction forces during long-term periods. With the Pedar Mobile system (using humidity-proof insoles, a zero setting after 1 h of usage, and the correction algorithm) we found errors of maximally 12% after a 7-h period (for the first peak in the ground reaction force) compared to Kistler ground reaction force data; the errors were only around 1–3% for the second peak in the ground reaction force. We believe that the present system can be used in a wide range of long-term measurement studies, such as studies on postoperative weight bearing measurements of orthopedic patients (e.g. total hip arthroplasty, osteotomies and fractures of lower extremity, cruciate ligament reconstruction).

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