

Temporal patterns of plantar pressures and lower-leg muscle activity during walking: effect of speed

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Abstract

Plantar pressure assessment is a tool useful for study of the gait cycle. In this study, we present a means of assessing the gait cycle using a temporal analysis of plantar pressures and lower-leg muscle activities. Plantar pressures and surface electromyography (EMG) of the tibialis anterior (TA) and medial gastrocnemius (MG) muscles were recorded as 19 men walked on a treadmill at seven speeds between 0.45 and 1.79 m/s. A typical ‘heel strike to toe off’ gait pattern was observed. Speed had minimal effects on the shapes of the muscle EMG root-mean-square—and plantar pressure–time curves except for the pressure–time curves in the heel and midfoot. A linear relationship was found between speed and peak pressures in the heel, medial forefoot, and toes; pressures in these regions increased by 91–289% going from 0.45 to 1.79 m/s. The temporal pressure changes in the forefoot and toes were paralleled by changes in MG muscle activity (i.e., cross-correlations of ≥ 0.90); TA muscle activity was not cross-correlated with the temporal pressure patterns in any region. However, the peak values of TA muscle activity were found to be highly correlated across speeds with peak pressures in the heel and toes (i.e., $r \geq 0.98$); similar high correlations were found between peak values of MG muscle activity and heel pressure. In summary, these data collected on able-bodied persons during motorized treadmill walking can be useful for comparison to those of patients undergoing treadmill evaluations for atypical gait cycle patterns and for tracking the progress of patients during gait rehabilitation.

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1. Introduction

The capacity to analyze gait quantitatively has been aided by recent improvements in the ability to assess plantar pressures during locomotion. The advent of new methodologies and increased computing power are mostly responsible for making plantar pressure assessment more accessible to the clinician. The technique is important to the clinician as it may be used for the assessment and management of lower-limb dysfunction resulting from injury or disease, in particular diabetic neuropathy [1–3].

We have recently begun assessing plantar pressures during the gait rehabilitation of patients following incomplete spinal cord injury (SCI). Gait assessment techniques commonly used for these patients are relatively limited and simple (e.g., determination of walking speed and total distance walked

[4,5]. It is thought that assessment of plantar pressures may provide a more comprehensive evaluation. During rehabilitation, patients with incomplete SCI can progress to supported walking with a consistent gait pattern but the patients may or may not walk using a ‘heel strike to toe off’ gait pattern typical of able-bodied persons under similar conditions [unpublished observations]. Assessment of peak plantar pressures alone is insufficient for detection of an atypical gait pattern as peak plantar pressures in these patients can approach those typical of able-bodied persons under similar walking conditions. We believe atypical gait patterns may be more readily detected if the plantar pressures are assessed temporally (i.e., determining how the pressures in a given plantar region rise and fall throughout the gait cycle and how that temporal pattern compares to those of other plantar regions). An analysis of the temporal patterns of plantar pressures during the gait cycle combined with assessments of lower-leg muscle activity patterns may be helpful in gauging the progress of patients during rehabilitation as well as the extent that their gait patterns return to those typical of able-bodied persons.

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Unfortunately, there are few data on able-bodied persons for comparison to those of patients. Only a few attempts have been made at describing the temporal patterns of plantar pressures during walking in able-bodied persons [6–8]. Two of these studies were case studies in nature describing the temporal patterns of plantar pressures during a single foot step made by an able-bodied person [6,8]. The data from the study of Duckworth et al. [7] do indicate how the plantar pressure–time relationships during a single foot step can vary among 50 able-bodied persons but these data were collected from subjects walking at differing speeds. It is generally accepted that walking speed affects plantar pressures. This premise is based on data from a few studies assessing peak plantar pressures over a narrow range of speeds [9–11] and from studies in which stride frequency was manipulated, presumably affecting walking speed but unfortunately speed was not reported [12,13]. Knowing the effect of speed on plantar pressure patterns is important because between-patient differences in walking speed are likely to be large and walking speed for a disabled individual is likely to increase as rehabilitation progresses.

The first objective of this study was to determine the temporal patterns of plantar pressures and lower-leg muscle activities over a wide range of walking speeds in able-bodied individuals with demographics similar to those incurring spinal cord injuries. It is intended that these data will be used for comparison to those of similar-aged patients with acute incomplete SCI or other lower-limb pathology. Furthermore, we describe in detail the analysis procedures applied so that similar analyses can be applied to other subject or patient populations. The second objective of this study was to determine the relationship of plantar pressure patterns during the gait cycle to those of lower-leg muscle activities. Knowing these relationships could be particularly useful for the clinician evaluating a patient's atypical plantar pressure patterns by helping to identify insufficient or inappropriate neuromuscular activities that might cause or predict the atypical pressure patterns.

2. Methods

2.1. Subjects

A convenience sample of 19 men from an university setting was used in this study. The subjects had no known lower-limb pathologies. Mean(\pm S.D.) age, height, and weight of the subjects were 29.0(\pm 4.7) years, 177.0(\pm 6.2) cm, and 87.5(\pm 21.3) kg, respectively. These demographics approximate those of people incurring SCI in the US as 82% of those injured are male with the average age being 32 years though 55% of those injured are between the ages of 16 and 30 [14]. Each subject gave informed consent prior to their participation in the study and the experimental protocol used in this study was approved by the institutional review boards.

2.2. Experimental protocol

After preparing a subject for measurement of plantar pressures and surface electromyography (EMG) (described below), the subject walked on a motorized treadmill on the level at 0.45, 0.67, 0.89, 1.12, 1.34, 1.56, and 1.79 m/s. The order of speeds was randomly assigned for each subject. Subjects walked for 1 min at a given speed prior to initiation of plantar pressure and EMG data collection. Data collection was initiated at the onset of stance phase of the right foot and continued for ten steps by each leg. Our rationale for using ten steps for data collection was based on the number (i.e., 9) previously determined to yield maximal reliability for plantar pressure measurement when using the same measurement system as employed in the present study [10]. The subjects spent \sim 10 min total on the treadmill.

2.3. Plantar pressure measurements

Plantar pressures during treadmill walking were measured using an EMED Pedar in-shoe plantar pressure measurement system (Novel Electronics, Inc.; St. Paul, MN). An insole in this system is placed between the shoe and sock and is made up of a matrix of 99 capacitance sensors with the area covered by a single sensor being \sim 1.5 cm². Pressure data from each sensor was sampled at 50 Hz. All insoles were calibrated before the study from 0 to 600 kPa according to manufacturer specifications. Each subject was provided with appropriately-sized Converse All Star Low M7652 court shoes and a pair of EMED insoles. The court shoes had been worn for 10–20 h prior to their use in this study. Nine different regions of the plantar surface of the foot were analyzed in this study. These regions were the lateral heel (LH), medial heel (MH), midfoot (MF), medial forefoot (MFF) consisting of the 1st metatarsal head, central forefoot (CFF) consisting of the 2nd and 3rd metatarsal heads, lateral forefoot (LFF) consisting of the 4th and 5th metatarsal heads, great toe (GT), middle toes (MT) consisting of the 2nd and 3rd toes, and lateral toes (LT) consisting of the 4th and 5th toes. Assignment of insole sensors to the nine plantar regions was based on foot tracings made directly onto the insoles. The median number of sensors per region was seven but ranged from five in both the GT and LT to 35 in the MF.

The rationale for our selection of the number and locations of the plantar regions was based on: (1) each region was to have functional characteristics differing from those of the other regions and (2) plantar pressures could be measured reliably in these regions. The regions we selected met these two criteria. First, all nine plantar regions had plantar pressure temporal patterns that were different from each other (see Figs. 1, 2 and 5). Second, reliability of the plantar pressure measurements was assessed in 16 subjects walking at their normal walking speed. Three trials were performed on one day and intraclass correlation coefficients (ICCs) were calculated for the peak pressure, time-to-peak pressure, and

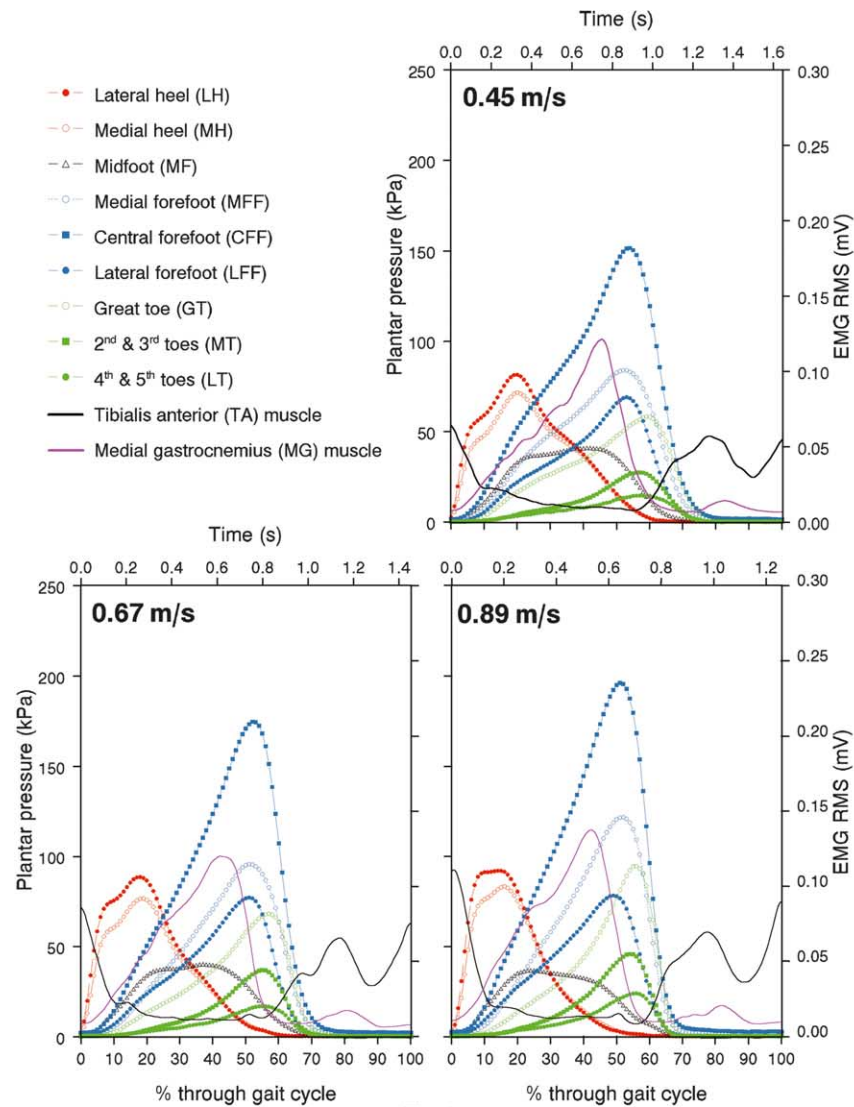


Fig. 1

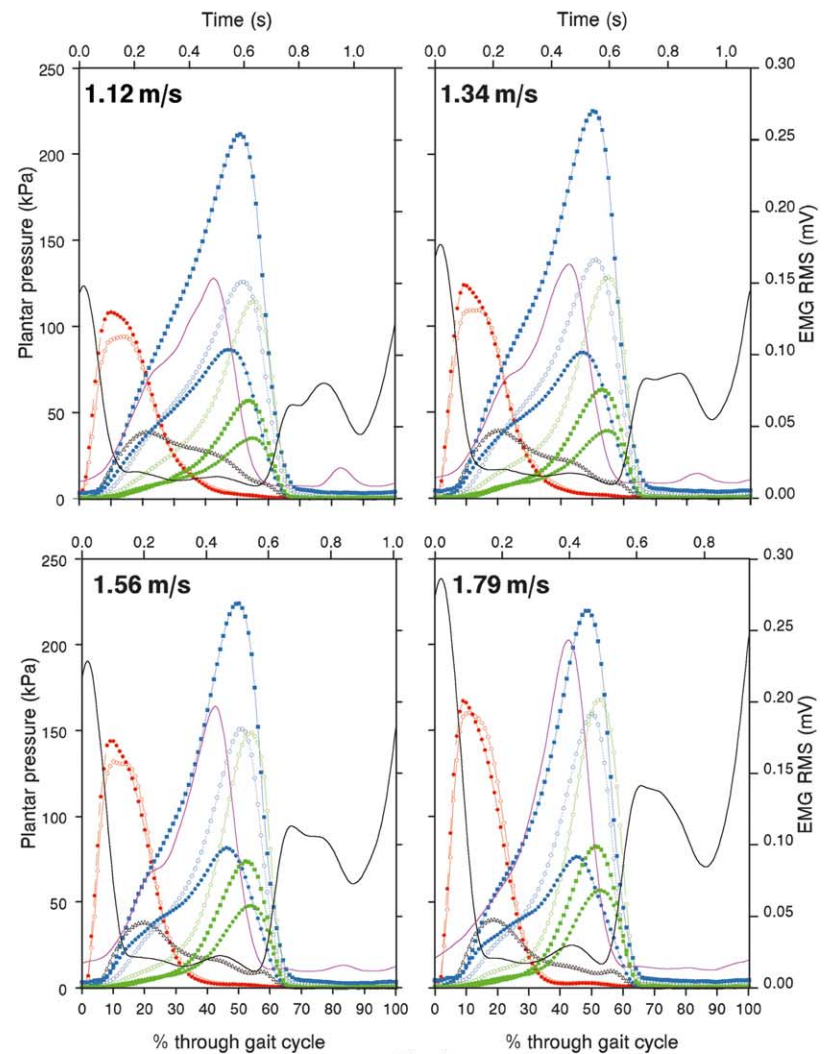


Fig. 2

Fig. 1. Mean plantar pressure– and muscle EMG RMS–time curves for walking speeds of 0.45–0.89 m/s. Error bars are not shown for the sake of clarity. In each plot, the length of top *x*-axis reflects the mean gait cycle duration for the 19 subjects.

Fig. 2. Mean plantar pressure– and muscle EMG RMS–time curves for walking speeds of 1.12–1.79 m/s. Symbols are the same as for Fig. 1.

pressure–time integral. The single lowest ICC for peak pressure or pressure–time integral in any plantar region in either foot was 0.97. The ICCs for time-to-peak pressure were slightly lower but the median ICC value of all 18 regions (i.e., two feet with nine regions each) was 0.97; the single lowest ICC was 0.87 and it occurred in the left MH.

2.4. Lower-leg muscle activity measurements

Surface EMG was used to assess the gait cycle activity patterns of the tibialis anterior (TA) and medial gastrocnemius (MG) muscles of the right leg. The TA and MG muscles were chosen because they are the principal dorsi- and plantar-flexor muscles, respectively. Placement of electrodes was as described by Warren et al. [15]. In brief, the leg's circumference was measured at one-third of the distance from the fibular head to the lateral malleolus. The pair of MG muscle electrodes was placed at this level after moving medially from the tibial crest by one-third of the leg's circumference. The pair of TA muscle electrodes was placed over the muscle belly at the same level along the length of the lower leg. The reference electrode was placed on the skin over the fibular head. After shaving, cleaning, and abrading the skin, 8-mm diameter Ag/AgCl electrodes (Grass-Telefactor model E55; West Warwick, RI) filled with conductive gel were placed on the skin. The center-to-center inter-electrode distance for a pair was 20 mm and a pair was aligned such that the line bisecting the two electrodes was intended to be parallel to the longitudinal axis of the underlying muscle fibers. Electrode impedance at 30 Hz was required to be below 5 k Ω and was usually \sim 1 k Ω .

The EMG electrode leads were inserted into bioamplifiers (Grass-Telefactor model P55). The EMG signals were amplified 1000 times and passed through bandpass filters with low and high cutoff frequencies of 10 and 1000 Hz, respectively. The amplified EMG signals were sampled at 5 kHz using a 16-bit A/D board (Keithley model KPCMCIA 16-AIAO-C; Cleveland, OH) and an 850 MHz laptop computer. EMG data acquisition was synchronized with that of the plantar pressures by using a synch out pulse from the EMED Pedar system to trigger data acquisition by the EMG system. Using the same electrode placements and a similar EMG data acquisition system, the reliability of EMG root-mean-square (RMS) determinations have been found to be good (i.e., ICCs of \geq 0.93 for within- and between-day retests) [15].

2.5. Data analysis procedures

Analyses were performed only on data from the right foot because lower-leg muscle activity patterns were acquired only for the right side. Initially, plantar pressure data were converted from Novel's proprietary format to ASCII format using PEDSTAN software (Novel Electronics, Inc.). All subsequent analyses were made using customized software written with TESTPOINT (version 4.0) (Capital Equipment Corp.;

Billerica, MA). Custom software was employed in lieu of software from Novel Electronics because we felt the custom software: (1) was less affected by sensor noise, (2) permitted a more objective and reliable determination of the times at which stance phase onset occurred, and (3) shortened considerably the time required for data analysis. The first step in a trial's data analysis was to determine the times at which stance phase onset occurred for the right foot. This was done by averaging pressures across all 99 sensors for each time point yielding a foot pressure versus time curve. The curve was subsequently filtered using a three-point median filter. The program then searched the foot pressure–time curve for stance phase onsets using the following criteria. First, onset of the stance phase had to occur at least 600 ms after the last. Second, the average foot pressure for the preceding 200 ms had to be less than 2 kPa (i.e., consistent with the foot being in swing phase). If these two criteria were met, the program searched for the first time point at which the derivative of the foot pressure–time curve exceeded 100 kPa/s for the six faster speeds and 60 kPa/s for the 0.45 m/s speed; for differentiation of the foot pressure–time curve, the data were first fit with a 4th order polynomial equation.

After determining the timing of the stance phase onsets, the analysis proceeded with step-averaging of the plantar pressure data from eight steps of the right foot. This process yielded for each sensor an averaged pressure–time curve that started at the onset of stance phase and ended with the end of swing phase. For a step's data to be used in the averaging, the step duration must not have exceeded the shortest step duration by more than 60 ms; usually at least eight of the ten steps met this criterion. For the longer duration steps meeting this criteria, the excess one to three data points were truncated before being averaged with the other steps. The pressure–time curves for all sensors within a region were averaged together to produce pressure–time curves for each of the nine plantar regions. In this averaging process, a sensor's pressure–time curve was weighted based on the relative area covered by the sensor in its region.

Analysis of a trial's EMG data started with generation of EMG RMS versus time curves for the two muscles. EMG RMS is a measure of overall muscle electrical activity and is determined by the number of motor units recruited as well as their firing rates. Every 5 ms throughout the ten-step data collection, RMS was calculated on a 50-ms window of data (i.e., 25 ms both before and after the data point of interest); any existing DC offset was removed prior to calculating the RMS. Using the stance phase onset times determined from the plantar pressure data analysis, the EMG RMS data were step-averaged using the same eight steps used in the plantar pressure data analysis.

Because of the considerable variation in step cycle duration among subjects (i.e., up to 58%), a subject's plantar pressure– and EMG RMS–time curves were converted from an absolute time scale to a relative one (i.e., as a percent through the gait cycle). For each individual, plantar pressure and EMG RMS values were calculated at 1%

intervals throughout the gait cycle using interpolation. After doing these conversions, the plantar pressure– and EMG RMS–time curves could be averaged across subjects.

2.6. Statistical analyses

Data in Section 3 are presented as means \pm S.E. except where noted. The within-subject effects of speed on peak muscle EMG RMS and peak plantar pressure in the nine plantar regions were analyzed using one-way repeated measures ANOVAs. When a significant effect of speed was found for peak muscle EMG RMS or peak pressure in a plantar region, within-subject contrasts were used to determine if the relationship was best described by a linear, quadratic, cubic, or higher order equation. Additionally, single degree-of-freedom contrasts were used to compare peak pressures at two or more speeds. Cross-correlation analyses were used to assess how well the within-subject temporal changes in plantar pressures during the gait cycle paralleled those of muscle EMG RMS. The within-subject relationships across speeds between peak plantar pressures and peak muscle EMG RMS values were analyzed using Pearson product-moment correlations. All statistical analyses were performed using spss (version 10.0.5) (SPSS, Inc.; Chicago, IL). An α level of 0.05 was used in all analyses.

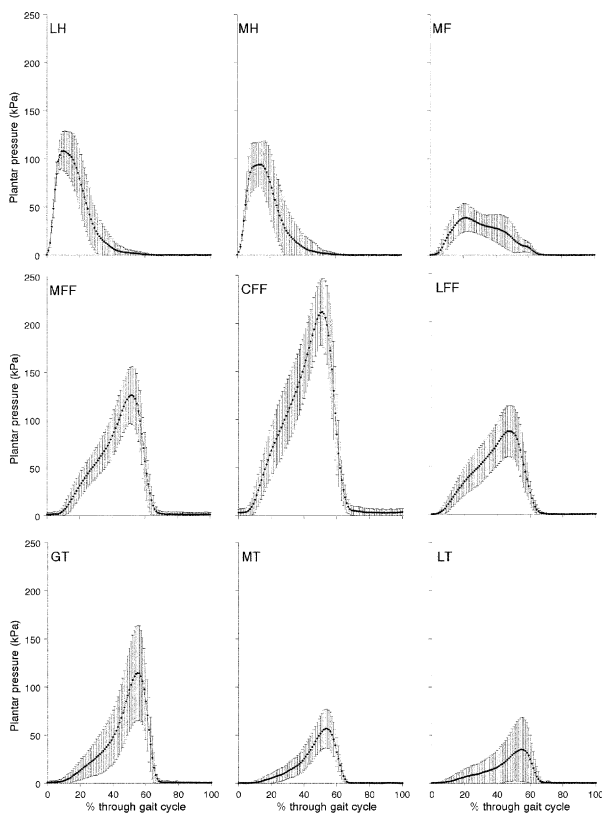


Fig. 3. Mean (\pm S.D.) plantar pressure–time curves for the 1.12 m/s walking speed.

3. Results

3.1. Gait pattern description

Figs. 1 and 2 depict the mean plantar pressure and muscle activity patterns while walking on the treadmill at speeds ranging from 0.45 to 1.79 m/s. For the sake of clarity, error bars have been omitted from these figures. Figs. 3 and 4 show the degree of variation among subjects in the plantar pressures and EMG RMS at the 1.12 m/s speed. The degree of variation at other speeds was proportionally similar. The greatest percentage variation in plantar pressures occurred in the LT with the standard deviation almost equaling the mean value, meaning that 68% of the sample had pressures in the plantar region ranging from almost zero to twice the mean. The relatively large variation observed among subjects in the EMG RMS–time curves is not unexpected given the unavoidable between-subject variability in electrode–tissue interface impedance and subcutaneous fat thickness.

A typical ‘heel strike to toe off’ gait pattern was observed during treadmill walking (Figs. 1 and 2). Early in the stance

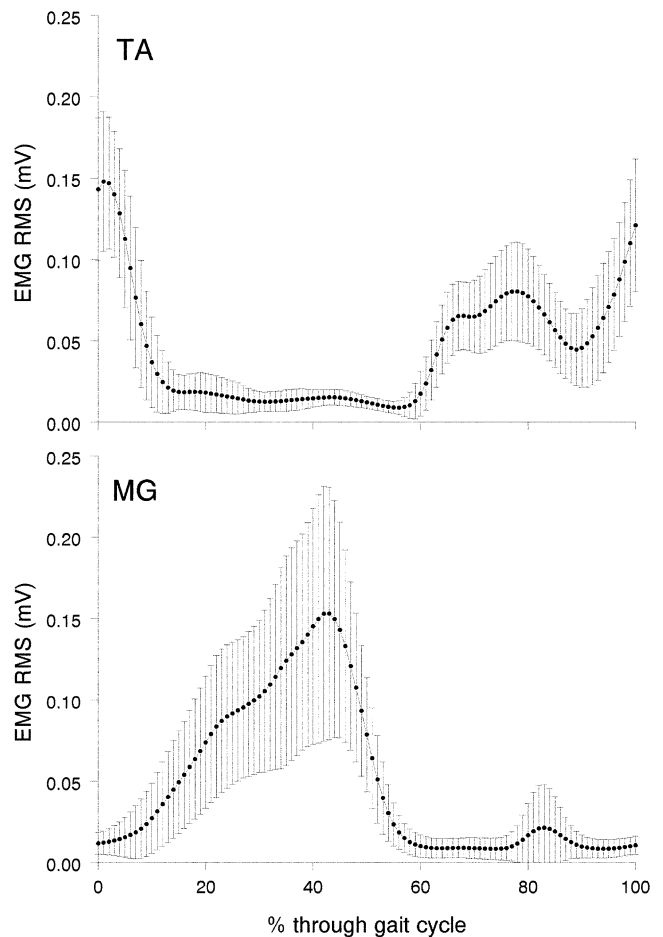


Fig. 4. Mean (\pm S.D.) muscle EMG RMS–time curves for the 1.12 m/s walking speed.

phase, pressures rose in the heel regions followed by a decline. As pressures in the heel declined, pressures in the forefoot and toe regions rose. As the stance phase ended, pressures in the forefoot and toe regions fell rapidly. TA muscle activity was high at heel strike, toe off, and during the swing phase; the highest activity occurred at the onset of the stance phase (i.e., 0–2% into the gait cycle). MG muscle activity was relatively low during the swing phase and at the onset of the stance phase but increased thereafter peaking at 42–46% into the gait cycle, i.e., slightly before the time forefoot pressures peaked. As determined from the plantar pressure data, the stance phase constituted ~79% of the gait cycle at the slowest speed but this proportion of the gait cycle decreased progressively as speed increased. At the highest speed, the stance phase constituted ~67% of the gait cycle. As discussed later, these estimates may be slight overestimates (i.e., by $\leq 2.5\%$).

3.2. Effect of speed on the shapes of the pressure- and EMG RMS-time curves

For all plantar regions with the exception of the heel and MF, the general shape of the pressure-time curves did not change appreciably with increasing speed (Figs. 1 and 2). As speed increased, pressures in the forefoot and toe regions did peak slightly earlier in the gait cycle. Peak pressures in the forefoot region occurred at 52–54% through the gait cycle at the slowest speed, decreasing to 45–50% into the gait cycle at the highest speed. The time of peak pressures in the toe regions decreased from 57 to 59% into the gait cycle at the slowest speed to 51–53% at the highest speed.

The pressure-time curves for the heel regions were relatively broad for the three slowest speeds. At the onset of the stance phase, heel pressures rose sharply until ~5–7% into

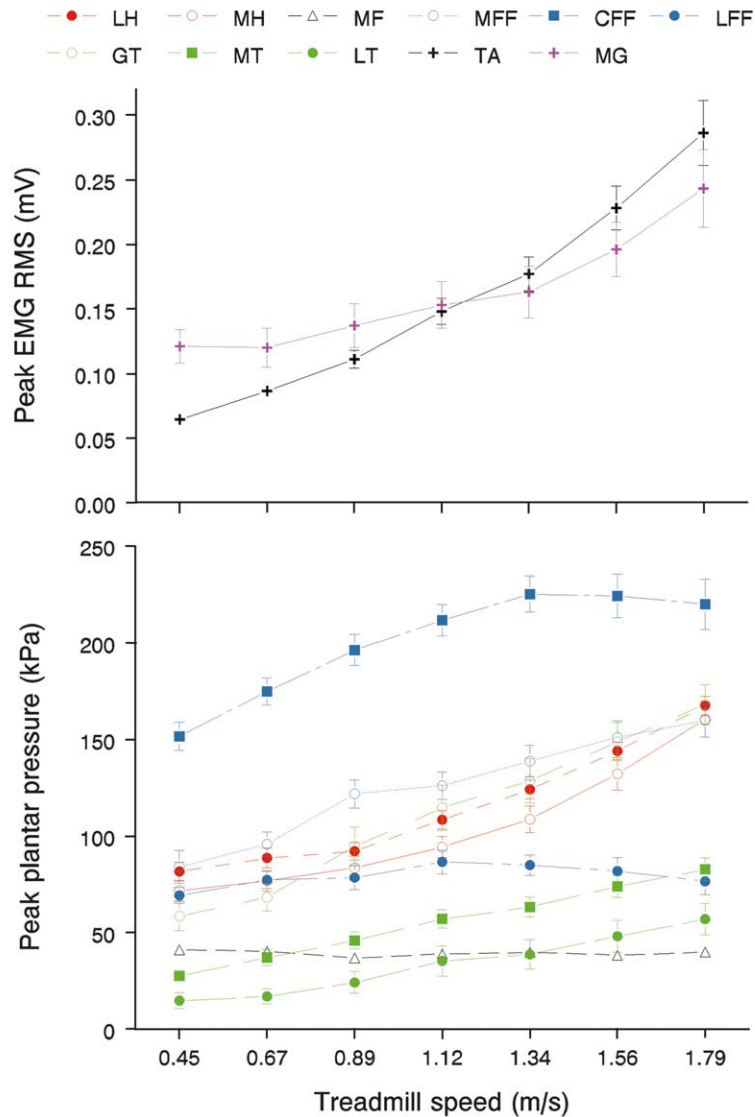


Fig. 5. Mean(\pm S.E.) peak values of the plantar pressure- and muscle EMG RMS-time curves as a function of walking speed.

the gait cycle and then more slowly until peak pressures occurred at 15–20% into the gait cycle. This two-phase increase in heel pressures was not observed at speeds >0.89 m/s. The declines in heel pressures after peaking were also relatively slow for the three slower speeds with pressures not reaching baseline until 55–60% into the gait cycle. However, as speed increased, the rate of decline in heel pressures was more rapid and at the highest speed, heel pressures had returned to baseline by ~37% into the gait cycle. The MF pressure–time curves for the 0.45–0.89 m/s speeds did not show distinct peaks as opposed to the higher speeds. At the slowest speed, peak MF pressures occurred at ~40% into the gait cycle but as speed exceeded 0.89 m/s, peak values occurred earlier, at ~20% into the gait cycle.

The general shapes of the two muscle EMG RMS–time curves were minimally affected by speed (Figs. 1 and 2). The most noticeable change was in the low-level burst of MG activity apparent midway through the swing phase (i.e., at 80–85% into the gait cycle). This activity increased in going from 0.45 to 1.12 m/s but decreased at higher speeds becoming nonexistent at the highest speed. This activity was quite variable among the subjects as reflected in Fig. 4.

3.3. Effect of speed on the peak values of plantar pressures and EMG RMS

Except for the MF ($P=0.33$), there were significant effects of speed on the peak values in the pressure–time curves for all plantar regions (Fig. 5). The effect was strong in the LH, MH, MFF, CFF, GT, MT and LT regions ($P < 0.001$). For the LFF, the effect was significant statistically ($P = 0.02$) but not practically as the variation in peak pressures across the 7 speeds was $\leq 25\%$ and there was no difference between the peak pressure at 0.45 m/s and that at 1.79 m/s. Except for the CFF and LFF regions, a linear equation best described the within-subject relationships between peak pressure and speed. The greatest percentage increases in peak pressure in going from the slowest to the fastest speed occurred in the toe regions (i.e., 188–289% increases); the heel regions exhibited the next largest increases (i.e., 105–124%) while the MFF exhibited a 91% increase. For the CFF, peak pressure increased linearly with speed up through 1.34 m/s, leveling off thereafter. CFF peak pressure increased by 49% in going from the slowest speed to the 1.34 m/s speed. Of the nine plantar regions, the CFF exhibited the highest peak pressure at all speeds.

There was a highly significant effect of speed on the peak values in the EMG RMS–time curves for both the TA and MG muscles ($P < 0.001$) (Fig. 5). Though the curves in Fig. 5 indicate some degree of curvilinearity, the within-subject data are better fit by linear relationships rather than quadratic ones. The percentage increases in peak EMG RMS in going from the slowest to the fastest speed were 101 and 344% for the MG and TA muscles, respectively.

3.4. Relationships between plantar pressures and muscle activities

As can be seen in Figs. 1 and 2, the temporal changes in the forefoot and toe region pressures over the gait cycle are paralleled fairly closely by the temporal changes in MG muscle activity. The cross-correlations between the forefoot plantar pressures and the MG muscle EMG RMS exceeded 0.90 at all speeds with the highest cross-correlations (i.e., 0.95–0.97) occurring at the highest speed. These strong cross-correlations are logical given the MG muscle's role as a plantar flexor muscle. The temporal changes in forefoot pressures did lag those changes observed in the MG muscle EMG RMS by 2–9% of the gait cycle. This lag should be expected when considering the time required to complete the muscle's excitation–contraction coupling process as well as that the elastic properties of force-transmitting structures in and outside of the body delay and slow the pressure changes as measured by the insole. The cross-correlations between the pressures in the toe regions and the MG muscle EMG RMS were slightly weaker (i.e., ≥ 0.87 at all speeds with the highest cross-correlations (i.e., 0.95) coming at the

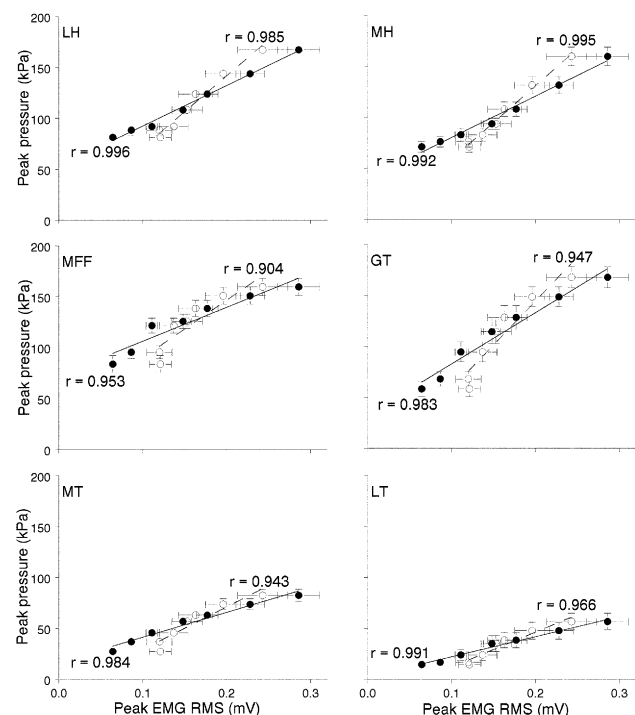


Fig. 6. Correlations across speeds of the peak values for muscle EMG RMS and plantar pressures. The filled and open circles reflect the data for the TA and MG muscles, respectively. The solid lines indicate the linear regression lines for the correlations between peak values of pressure and the TA muscle EMG RMS; the dashed lines indicate regression lines for the MG muscle. The correlation coefficients for the TA and MG muscles are indicated at the lower left and upper right, respectively, in each plot. All significant correlations between peak values of pressure and EMG RMS are shown except for that between CFF pressures and TA muscle EMG RMS; in this case, the correlation was relatively weak (i.e., $r=0.83$; $P=0.02$).

highest speed). The temporal changes in the toe region pressures lagged those of the MG muscle EMG RMS by 9–13% of the gait cycle. There were no strong cross-correlations found between the temporal changes in TA muscle activity and the temporal changes in pressure observed for any plantar region.

Based on the cross-correlations just presented, one might predict that across speeds the peak values of MG muscle EMG RMS would be highly correlated with the peak pressures in the forefoot and toe regions. This was not the case, at least not for the forefoot regions as only the MFF exhibited peak pressures that were significantly correlated with peak values of MG muscle EMG RMS ($r = 0.90$; $P = 0.005$) (Fig. 6). All three toe regions did exhibit peak pressures that were correlated with peak values of MG muscle EMG RMS ($r = 0.94$ – 0.97 ; $P < 0.001$). Surprisingly, the peak values of MG muscle EMG RMS were highly correlated with the peak pressures observed in the heel regions ($r = 0.99$ – 1.00 ; $P < 0.001$). A direct cause-and-effect relationship is not likely to explain this observation. Though no strong cross-correlations were found to exist between TA muscle activity and the pressures in any plantar region, the peak values of TA muscle activity were found to be highly correlated across speeds with the peak pressures in the two heel and three toe regions ($r = 0.98$ – 1.00 ; $P < 0.001$) (Fig. 6). Again, a direct cause-and-effect relationship is not likely to explain these observations but knowledge of these relationships may be useful in predicting insufficient or inappropriate neuromuscular activities from atypical plantar pressure patterns or vice versa.

4. Discussion

These are the first data we are aware of that describe in detail the temporal patterns of plantar pressures during motorized treadmill walking by able-bodied persons and how walking speed affects these patterns. Furthermore, these data are the first relating ankle dorsi- and plantar-flexor muscle activities to the rise and fall of plantar pressures during the gait cycle as well as to the peak pressures themselves. We believe that these data and the analysis procedures developed for the study can assist clinicians in identifying and perhaps treating atypical gait patterns. Simply analyzing the so-called standard plantar pressure measures (e.g., peak pressure and time-to-peak pressure) may or may not be helpful in the detection of a patient's atypical gait. It should be easy to visualize a patient with an atypical gait who exhibits an abnormal rise or fall in pressures in one or more plantar regions yet has relatively normal peak pressures and times-to-peak pressure. Visual inspection of such a patient's plantar pressure- and EMG RMS-time curves and subsequent comparison to the curves of persons without lower-limb pathology (e.g., like those in Figs. 1 and 2) would be helpful in identification of the atypical gait characteristics. For persons wishing to perform these types of

plantar pressure- and EMG RMS-time analyses on Novel EMED Pedar data, the analysis software used in this study is available upon request from the corresponding author.

The present study's findings regarding the effect of walking speed on plantar pressures are in general agreement with those of earlier studies. The few published studies investigating this have focused on how peak pressures and to a lesser extent how times-to-peak pressure vary over either two [11] or three [9,10] speeds. Measurements in one of these studies were made while the subjects walked barefoot on a pressure-sensitive mat [9] whereas the other two studies used the same plantar pressure measurement system as used in the present study [10,11]. In agreement with the present study, these three prior studies also failed to report a significant effect of speed on peak pressure in the MF. Also in agreement with the present study, these three studies reported no increase in LFF peak pressure as speed increased; Rosenbaum et al. [9] actually reported a 35% decrease in the 5th metatarsus region as speed increased from 0.83 to 1.67 m/s. The present study and all three studies found heel peak pressures to increase significantly as speed increased. Rosenbaum et al. [9] and the present study also reported peak pressures in the MFF, central forefoot, and all toe regions to increase as speed increased; Kernozek et al. [10] reported increases in the toes but not in the medial or central forefoot. As mentioned previously, peak pressures in the central forefoot increased only over speeds at ≤ 1.34 m/s for the present study.

For the plantar regions demonstrating increases in peak pressure as speed increased, the percentage increases varied considerably among studies. The studies of Drerup et al. [11] and Kernozek et al. [10] generally reported lesser percentage increases when compared to the present study over a similar range of speeds. On the other hand, the present study and Rosenbaum et al. [9] found similar percentage increases when compared over a similar range of speeds. The explanation for this variation among studies is not readily apparent but is probably due to variation among studies in the experimental protocol (e.g., walking barefoot [9] versus in shoes ([10,11], present study) or in differing footwear ([10,11], present study)) and/or the plantar pressure measurement methodology (e.g., using the peak pressure value for a region as the highest value from any sensor in the region [9–11] versus using the peak value from the pressure-time curve produced after averaging across all sensors in a region and subsequently after averaging across all subjects (present study)).

Though we believe the data from the present study can be useful to the clinician in identifying atypical gait due to lower-limb pathology, we must point out that these data have limitations. First, caution should be applied when comparing these data to those collected from overground walking trials as it has been reported that gait characteristics can differ between treadmill and overground walking [16–18]. We opted to collect data on the treadmill because patients with SCI are commonly evaluated and/or

rehabilitated using a treadmill. Second, we did not assess the effect of foot type (e.g., high, normal, or low arch) and it is possible that foot type may alter the temporal plantar pressure or muscle activity patterns along with their speed dependence. However, Rosenbaum et al. [9] previously reported foot type to not affect how peak plantar pressures change as walking speed increases from 0.83 to 1.67 m/s. Furthermore, we thought that any alterations in the temporal patterns induced by foot type would be relatively small compared to those induced by lower-limb pathology and thus varying foot type among patients undergoing assessment would not impair our ability to detect an atypical gait pattern. Finally, the present study's measurements were made with the subjects wearing a basic athletic shoe and it is likely that wearing a more cushioned or supportive shoe would alter the plantar pressure- and EMG RMS-time curves.

We are also aware of modifications that would improve the temporal pattern analysis procedures introduced in this study. First, the Novel EMED Pedar plantar pressure measurement system permits disabling of sensors in order to increase the sampling frequency of the remaining sensors. If the sensors for the left foot had been disabled, we could have achieved a twofold better temporal resolution for the right foot. Second, for each walking trial, we averaged the pressure- and EMG RMS-time curves across eight steps. The averaging method that we used works perfectly only if all steps are of the same duration. Because this does not occur often, we decided to exclude foot steps more than 60 ms longer than the shortest. For foot steps 20–60 ms longer than the shortest foot step, the so-called extra data points (1–3 for the plantar pressure-time curves and 4–12 for the EMG-time curves) at the end were not used. This truncation of data produced a shortcoming only noticeable in the EMG RMS-time curve for the TA muscle (e.g., the 1.79 m/s speed in Fig. 2) in that the EMG RMS at 0% into the gait cycle did not match up with the value at 100%, when in fact they should. This shortcoming did not affect the plantar pressure-time curves other than produce a slight overestimation (i.e., $\leq 2.5\%$) of the stance phase duration and other timing measures. We now realize that we could have overcome this problem by converting the time scale for each foot step to a relative one (i.e., as a percent through the gait cycle) before averaging across foot steps and then using interpolation to predict plantar pressure values for integer values between 0 and 100% throughout the gait cycle.

In summary, we have presented in detail a means for assessing the gait cycle using a temporal analysis of plantar pressures and lower-leg muscle activities. The data collected on able-bodied men during motorized treadmill walking can be useful for comparison to those of patients undergoing treadmill evaluations for atypical gait cycle patterns and for tracking the progress of patients during gait rehabilitation. Future studies should be directed towards assessing how gender, differing footwear, and walking sur-

face affect the temporal patterns of plantar pressures and lower-leg muscle activities as well as their speed dependence. In these studies, it would also be prudent to control for or at least monitor other factors that could affect the temporal patterns (e.g., foot type, foot progression angle, plantar tissue characteristics, joint mobility, abnormal bony foot structures, femoral anteversion/retroversion, tibial torsion).

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