Surface muscle pressure as a measure of active and passive behavior of muscles during gait

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1. Introduction

Recording muscle activities with surface electromyography (SEMG) is essential to many applications such as assessment of injury or disease, control of external devices, biofeedback, and human performance measurements. SEMG profiles of specific muscles inform treatment decisions and measure outcomes in many biomechanical and neuromuscular conditions, such as cerebral palsy, spinal cord injury, Parkinson’s disease, arthritis and leg injuries [1,2]. SEMG signals are useful for biofeedback training and to control assistive devices, and they have become a valuable tool for sports and athletic training, monitoring muscle fatigue, and ergonomic studies [3].

SEMGs can accurately record the electrical activities of specific muscle fibers, and provide some estimate of their contractile forces [4], but are insensitive to the passive tensions experienced by and subsequent work produced by the musculo-tendinous unit (MTU).

While surface electromyography (SEMG) can accurately register electrical activity of muscles during gait, there are no methods to estimate muscular force non-invasively. To better understand the mechanical behavior of muscle, we evaluated surface muscle pressure (SMP) in conjunction with SEMG. Changes in anterior thigh radial pressure during isometric contractions and gait were registered by pressure sensors on the limb. During isometric knee extensions by a single subject, SMP waveforms correlated well with SEMG (r = 0.97), and SEMG onsets preceded those of SMP by 35–40 ms. SMP and SEMG signals were simultaneously recorded from the quadriceps of 10 healthy subjects during gait at speeds of 0.4, 0.8, 1.1, 1.4 and 2.2 m/s. Muscle activity onset and cessation times were objectively determined for both modalities, and results showed high intra-class correlations. SMP waveforms were highly consistent from stride to stride, while SEMG waveforms varied widely. SMP waveforms were typically brief, while SMP waveforms tended to be biphasic and outlasted the SEMG by approximately 40% of gait cycle at all speeds. These results are consistent with mechanical models of muscle, and demonstrate the use of SMP to estimate the timing of knee extensor muscle stiffness during gait.

These forces are generated by elastic stretching of the MTU during joint motions, and can play major roles in lower limb joint torques and energy usage, during walking [5]. Assessment of these passive and active forces of leg muscles during gait has been done using intramuscular sonography [6]. For example, real-time ultrasonic scanning of the gastrocnemius revealed a 7 mm stretch of the tendon during stance phase of gait, followed by recoil at push-off. These experimental studies, as well as dynamic modeling, have suggested that the MTU and the fascicles of bi-articular muscles, such as the rectus femoris, undergo significant stretching during the stance phase of gait, producing tension that returns energy to the working joints [5].

This study presents a method to detect the timing and estimate the relative amplitude of active and passive behaviors of working muscles, using dynamic surface muscle pressure (SMP) recordings of the anterior compartment of the thigh during gait via sensors placed over the three superficial heads of the quadriceps. SMP measures the radially oriented pressures at the surface of the limb that are the result of muscle shape changes during activation. Our previous work applied SMP to forearm muscles, where it accurately estimated grip force [10] and encoded hand volitions with high resolution [11].

Several other studies have established a relationship between SMP and muscle activity. In particular, it is known that surface...
muscle pressure correlates well with intramuscular pressure (IMP) \[7,8,12\], and the latter reflects axial muscle force during voluntary contraction, as well as during passive stretching \[13\]. The linear relationship between SMP and IMP was shown in a study using pressure sensor catheters inserted into the biceps, along with mechanomyography and sphyngomanometry \[12\]. Further studies on the human arm and leg muscles, both in vivo for humans and ex vivo for rabbits \[9,14–17\], showed that IMP correlates well with axial muscle pressure in a nearly linear fashion over a large range of effort. During isometric leg contractions, IMP of the \textit{vastus medialis} increased linearly with force up to the maximum voluntary contraction (MVC) \[15\]. Moreover, IMP increased with passive stretching of the \textit{anterior tibialis} \[16\]. During walking, the \textit{anterior tibialis} experiences a large (35 mm Hg) increase in IMP during foot raise, which continues during early stance phase, long after SEMG is silent \[14\]. These studies thus clearly suggest that SMP can accurately reflect both active and passive forces in muscles.

Herein we measured SMP of the anterior thigh simultaneously with SEMG during gait at a range of speeds, in order to (1) evaluate the consistency of SMP in detecting muscle activity onset relative the SEMG and (2) gain insight into the mechanical forces generated within the limb during gait. In the present study, \textit{quadriceps} activity was recorded via both SMP and SEMG modalities during gait at speeds ranging from 0.4 m/s to 2.2 m/s.

2. Methods

2.1. Human subject protection

This study complied with the principles of the Declaration of Helsinki and was approved by the Creighton University Institutional Review Board. All subject volunteers read and stated they understood the informed consent document. The researcher assured their understanding by verbally summarizing the project and the associated risks in addition to having the subject restate the tasks and risk. The researcher answered all subject questions before the subject signed the informed consent document.

2.1.2. Subjects

Ten healthy young adults (2 males and 8 females) with no significant limitations or active pathology in their lower extremities were recruited from a sample of convenience for the ambulatory trials. Mean age was 23.3 years (SD 1.25), mean male subject body mass was 76.5 kg (SD 15.1), and mean female subject body mass was 72.5 kg (SD 14.9). Additionally, a single male subject, aged 53, was tested on an isometric task. The dominant leg (leg preferred to kick a ball) was tested, which was the right leg for all subjects in our pool.

2.2. SEMG recording

After removing hair and cleansing the skin with alcohol at application sites, pairs of SEMG surface electrodes were applied to the skin of the thigh parallel to the primary axis of the underlying muscle fibers, according to standard procedure \[12\]. Three pairs were applied to each of three regions of the \textit{quadriceps} muscle (\textit{vastus lateralis}, \textit{rectus femoris} and \textit{vastus medialis}). Electrode pickups were 1 cm in diameter and spaced 2.5 cm apart on center (ConMed Huggables pediatric ECG electrodes #1620, Utica, NY, USA). Short shielded connector wires were routed to the lateral thigh and attached to preamplifier units. The signals were routed by fiber optic cable to the main SEMG amplifier (Motion Lab Systems MA300, Baton Rouge, LA, USA). The amplified analog signals were digitally sampled at 100 Hz, as has been used previously \[17,18\]. Signals were high-pass filtered with a 4th order, 4 Hz cutoff Butterworth filter.

2.3. Surface muscle pressure

Eight pressure sensors were developed using force sensitive resistors (Interlink Electronics, Carpinteria, CA, USA) and were mounted on foam material arranged in three columns (three lateral, two midline and three medial), to overlie the \textit{vastus lateralis}, \textit{rectus femoris} and the \textit{vastus medialis} heads of the quadriceps in an adult thigh, as shown in Fig. 1. Pressure sensors were located near to but not in contact with the SEMG electrodes or the connector wires and held in place with self-adherent circumferential elastic wrap (3M Coban Self-adherent Wrap, 3M Corporation, St. Paul, MN, USA). The application provided a baseline static pressure with a uniform, comfortable fit while allowing for detection of local positive as well as negative pressure changes. Voltage output from each sensor was obtained from a half-bridge circuit with the force sensitive resistor and a 10 kΩ fixed resistor; no attempt was made to record absolute pressure. Sensor outputs to calibrated pressures were linear between 0 and 20 Hz, and were suitable for registering myokinetic activity on the surface of the limb \[11\]. SMP signals were sampled synchronously with SEMG signals at 100 Hz, as above. Low-pass filters were used as described above for the SEMG.

2.4. Footswitches

Footswitches were taped to the bottom of the subject’s shoes, one on each heel lateral to the midline of the foot and one on each
forefoot beneath the head of the first metatarsal. Footswitches were interfaced to the computer through the Motion Lab Systems ampli-
fier to provide fiducial points for the cyclic functional activities. Two
tracks of footswitch data (heel and toe) were captured at 100 Hz as
an independent reference of temporal landmarks.

2.5. Isometric protocol

A single male subject, 53 years old with no significant lim-
itations, and who had previous experience with isometric and
isokinetic knee testing, was seated in an isokinetic dynamometer
(Biodex Medical Systems, System 3, software version 2.15) with hip
and knee both positioned at 60° flexion. Following manufacturer’s
recommendations, restraining belts were placed across the right
thigh at the waist, and crossed over the thorax. All isometric data
were collected from the subject’s dominant right leg.
Approximately 10–15 s of SEMG and SMP data were simultane-
ously collected for two isometric trials. Each maximum voluntary
isometric extension was held at for about 5 s, at a maintained torque
of approximately 115 ft lbs with a 2 min rest between contractions.

2.6. Ambulatory protocol

All trials were performed on a clinical treadmill (RTM400, Biodex
Medical Systems, Inc., Shirley, NY, USA). The subjects ambulated
continuously with approximately 1 min of warm up followed by
20–30 s data collection epochs over five speeds: 0.4 m/s, 0.8 m/s,
1.1 m/s, 1.4 m/s, and 2.2 m/s. Speeds were presented from slowest
to fastest in all subjects. All subjects transitioned from a walking
gait to a running pattern at 2.2 m/s. Subjects were allowed 30–60 s
after each change of speed to stabilize their gait pattern before data
capture. The treadmill speeds were representative of gait speeds
ranging from very slow walking (typical of severe gait impairment)
to moderate jogging.

2.7. Data analysis

The final SEMG and SMP signals used for analysis were composites
obtained by summing all corresponding sensors on the anterior
thigh muscles. Raw signals were demeaned, full-wave rectified,
summed across all sensors, and then smoothed. Signals from rectus
femoris and the vasti were summed together because of the abil-
y of SMP to register activity in proximal and distal sensors; the
resultant records were reflective of activity within an anatomic
compartment and made no attempt to differentiate between the
heads of the quadriceps. Composite signals were smoothed using a
5-point moving average. In this way, conditioned data tracks from
both measurement modalities reflected identical post-hoc signal
processing. Note that similar processing was used for the isometric
protocol, except low pass filtering was accomplished using either
a 7-point moving average or a 4th order Butterworth filter with a
cutoff of 20 Hz.

Gait cycles (strides) were delineated as the period between two
consecutive heel strikes, and all data were time normalized to per-
cent of cycle, as defined by footswitch data. To systematize timing
of muscle onset and cessation during gait, SEMG onset and ces-
sation times were found by setting a threshold above (or below)
‘baseline’ activity, defined as the standard deviation of the signal
between 40% and 60% of gait cycle. Both the vasti and the rectus
femoris exhibited minimal SEMG activity during this period, which
Corresponded with the late stance to pre-swing phases of the gait
cycle[13]. Within each gait cycle, SEMG onset times were defined as
the time-point, i, at which the ith SEMG track, \( E_i \), exceeded a thresh-
old value of \( \bar{\sigma} \cdot \sigma \), where \( \bar{\sigma} \) denotes the mean over all gait cycles of
the standard deviation of 20 time-samples within the quiet period
of quadriceps activity. \( \bar{\sigma} \) is defined as

\[
\bar{\sigma} = \frac{1}{N_{GC}} \sum_{i=1}^{N_{GC}} \frac{1}{10} \sum_{j=40}^{60} (E_{i,j} - \bar{E}_{40-60})^2.
\]

where \( N_{GC} \) is the number of gait cycles, where \( E_{i,j} \) is the SEMG value
of the ith stride at the jth percent of gait cycle, \( \bar{E}_{40-60} \) is the mean
of values from the ith stride between 40% and 60% of gait cycle.
Cessation times for SEMG were defined as the point at which the
recorded value is sustained below a value set at twice the threshold
defined by Equation 1. Spurious peaks in the heel switch record
were manually eliminated to ensure the detection of correct stride
time. Additionally, the automatic detection of fiduciary markers
was confirmed visually, in accordance with accepted methods[14].

SMP onsets were defined as the maximum of the time-derivative
of each temporally normalized SMP waveform. Derivatives were
estimated using a 3rd-order spline. SMP cessation was determined
as the minimum of the time derivative.

In the present experimental design, there are two indepen-
dent raters of muscle activity: SMP and SEMG. Comparisons were

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Fig. 2. SMP and SEMG records from isometric contractions during knee extension, comparing Butterworth low-pass filter (bottom panel) and moving average filter (top
traces) to raw signals (lower panel). (A) SEMG records from quadriceps during isometric knee extension. (B) Simultaneous SMP records from anterior thigh during extension.
(C) Superimposed and expanded SEMG and SMP signals from panels (A) and (B) during knee extension.
therefore made on the basis of intraclass correlation coefficients (ICC(2,1)), which are equivalent to a two-way ANOVA. ICC values were calculated to determine the inter-rater differences in onset and cessation timing using the method of McGraw and Wong [15]:

$$\text{ICC}(2, 1) = \frac{\text{MSR} - \text{MSE}}{\text{MSR} + \text{MSE}}$$

(2)

where MSR and MSE are the mean squares of row variables and error, respectively, obtained from analysis of variance. ICC values range from 0 to 1, representing completely uncorrelated and identical data sets, respectively.

Waveform variability from stride to stride was quantified as the variance ratio (VR) [16]:

$$\text{VR} = \frac{\sum_{i=1}^{k} \sum_{j=1}^{n} (X_{ij} - \bar{X}_j)^2 / (k N_{GC} - 1)}{\sum_{i=1}^{k} \sum_{j=1}^{n} (X_{ij} - \bar{X})^2 / (k N_{GC} - 1) - 1}$$

(3)

where $k$ is the number of data points in the stride, $N_{GC}$ is the number of gait cycles being analyzed, $X_{ij}$ is a specific data point, $\bar{X}_j$ is the ensemble average at time $i$, and $\bar{X}$ is the global average. VR is a two-way statistic reflecting waveform variance among strides, with 0 indicating all identical signals and 1 representing a series of random waves.

3. Results

3.1. Isometric waveform comparison

Comparison of the isometric quadriceps SEMG signal with the simultaneous SMP signal from the anterior thigh is shown in Fig. 2 for two filter types. Top traces of panels A and B were processed with the moving average filter, with the raw signals immediately below; bottom traces were processed with the Butterworth filter, and the
same raw signals are also shown. Both the SMP and SEMG signals were quiet prior to onset of the extensor contraction and rose with a time constant of about 70 ms to a relatively stable but oscillatory level. Superimposition of the processed signals on an expanded scale (Fig. 2C) shows that SMP closely followed the onset of SEMG after a delay of 35–40 ms, but oscillated less regardless of filter type. Cross-correlations between the entire waveforms were high, with $r = 0.92 \pm 0.001$ and $r = 0.97 \pm 0.001$ using the Butterworth filter and the moving average filter, respectively.

3.2. Gait waveform comparison

The grouped SEMG and SMP signals from the anterior thigh are shown in Fig. 3, which includes representative gait cycles from individual strides by a single subject at three different ambulation speeds. At the slowest speed, 0.4 m/s, SEMG records are difficult to visually interpret due to their high variability. At 1.1 and 2.2 m/s, the SEMG activation pattern is more apparent in the raw signals, whereby a distinct, brief peak is observed to begin at or slightly before heel strike, as expected. SMP traces were more repeatable at each speed and exhibited a temporal pattern similar to that of SEMG around heel strike. Unlike SEMG, which was briefly active around heel strike, the SMP persisted during stance phase, and often produced a prominent second peak near 50% of cycle, as clearly seen at the 1.1 m/s speed. SMP signals for all subjects, across all speeds, were above threshold for 66.4 ± 4.5% of the gait cycle.

The relative variability of the waveforms was quantified with VRs for all ambulatory trials, as shown in Fig. 4, where SEMG is seen to be 5–36 times more variable than SMP. At the slowest speed, 0.4 m/s, the SEMG VR of 0.97 indicates that the signals from stride to stride were almost completely unrelated to each other. As speed increased, VR of SEMG decreased monotonically from 0.97 to 0.29. In contrast, VR of SMP was relatively low at 0.4 m/s (VR = 0.1), and decreased to 0.02–0.06 at faster speeds. Note that the SMP minimum VR occurred at speeds between 1.0 and 1.5 m/s, near the preferred walking speed.

3.3. Activity timing during gait

SMP and SEMG signals at five speeds were processed to identify temporal landmarks of onset and cessation as illustrated in Fig. 5, for a single subject ambulating at 1.4 m/s. The On/Off spikes below each trace in Fig. 5 show the onset (upward) and cessation (downward) of activity during each stride. Toe-off in all cases occurred between 40% and 60% of the gait cycle. Minor SEMG activity near 50% of gait cycle rarely exceeded threshold, resulting in infrequent occurrences of spuriously high peaks that were discounted. Note that average onset times for SMP consistently lagged those determined by SEMG, and increased from 3% of cycle at the slowest speed to 12.5% at the fastest as shown in Table 1, for all subjects. These lags corresponded to average time delays ranging from 54 to 94 ms. Onset timing correlation between the two modalities, as measured by ICC, ranged from 0.70 (SD 0.14) at the slowest speed (0.4 m/s) to 0.83 (SD 0.11) at the fastest speed (2.2 m/s).

SMP suprathreshold activity persisted for a much longer fraction of the gait cycle, compared with SEMG, as seen in Fig. 5 and Table 1. Average duration of SEMG signal ranged from 14.8% to 26.7% of cycle, with the briefest signals occurring at the intermediate speeds of 1.1 and 1.4 m/s. In contrast, SMP persisted for 70.2% of cycle at the slowest speed, and monotonically decreased with speed, to 56.2% at the fastest. These delays from cessation of the SEMG to SMP signals corresponded to a range of 805 ms at the slowest speed to 234 ms at the fastest. The ICCs for cessation (Table 1) ranged from 0.99 ± 0.009 at 0.4 m/s, to 0.92 ± 0.075 at 2.2 m/s.
Onset and cessation timing of SEMG and SMP as percentage of gait cycle (mean ± st. dev.), and ICC values by speed across subjects (N = 10).

<table>
<thead>
<tr>
<th></th>
<th>0.4 m/s</th>
<th>0.8 m/s</th>
<th>1.1 m/s</th>
<th>1.4 m/s</th>
<th>2.2 m/s</th>
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</thead>
<tbody>
<tr>
<td><strong>Onset</strong></td>
<td></td>
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<td></td>
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<tr>
<td>SEMG (%)</td>
<td>5.4 ± 8.7</td>
<td>-0.2 ± 4.6</td>
<td>-0.5 ± 4.2</td>
<td>-6.0 ± 3.9</td>
<td>-6.0 ± 4.5</td>
</tr>
<tr>
<td>SMP (%)</td>
<td>8.4 ± 7.4</td>
<td>7.7 ± 5.4</td>
<td>7.5 ± 5.0</td>
<td>3.0 ± 4.5</td>
<td>6.5 ± 4.2</td>
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<tr>
<td>Avg. lag (ms)</td>
<td>55</td>
<td>46</td>
<td>90</td>
<td>95</td>
<td>94</td>
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<tr>
<td>ICC</td>
<td>0.70 ± 0.14</td>
<td>0.75 ± 0.10</td>
<td>0.79 ± 0.12</td>
<td>0.81 ± 0.16</td>
<td>0.83 ± 0.11</td>
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<td><strong>Cessation</strong></td>
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<td>SEMG (%)</td>
<td>25.9 ± 8.8</td>
<td>22.6 ± 3.7</td>
<td>20.7 ± 4.5</td>
<td>14.8 ± 4.3</td>
<td>26.7 ± 4.7</td>
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<tr>
<td>SMP (%)</td>
<td>70.3 ± 5.3</td>
<td>66.4 ± 3.0</td>
<td>65.4 ± 5.7</td>
<td>59.8 ± 8.0</td>
<td>56.2 ± 10.4</td>
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<tr>
<td>Avg. lag (ms)</td>
<td>805</td>
<td>581</td>
<td>509</td>
<td>477</td>
<td>234</td>
</tr>
<tr>
<td>ICC</td>
<td>0.99 ± 0.009</td>
<td>0.99 ± 0.003</td>
<td>0.98 ± 0.021</td>
<td>0.97 ± 0.034</td>
<td>0.92 ± 0.075</td>
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4. Discussion

4.1. Validity of the study

We studied 10 subjects ambulating on a treadmill at speeds that ranged from slow walking (0.4 m/s) to jogging (2.2 m/s), while SMP and SEMG sensors simultaneously monitored the quadriceps; we also analyzed isometric records from a single subject. Systematic temporal comparisons between the two modalities, in terms of activity onset and cessation, were done using standard automated methods for detection of muscle onset [24]. SEMG temporal landmarks were demarcated by applying the standard threshold value of 2 · σ. We based σ on the period between 40% and 60% of gait cycle, when SEMG amplitude is minimal, although it is not entirely quiet, as seen from Fig. 5, in accordance with previous observation [13]. Individual sensor records were analyzed to confirm that the baseline derivation from 40% to 60% of gait cycle was extracted from a quiet period. For SMP temporal measurements, we based threshold on the derivative, since a quiet period was undefined. The 100 Hz data sampling rate was adequate to accurately identify and compare onset and cessation within the 101-point scale of our gait-cycle normalized data, as demonstrated previously [17,18].

Our intent was to capture and compare muscle activity profiles within an anatomic compartment as opposed to a more traditional EMG analysis that might differentiate between muscle heads. Since SEMG and SMP are independent raters of muscle activity, they were compared using a two-way ANOVA, which yielded the intraclass correlation coefficients listed in Table 1. Overall waveform variability was assessed by variance ratios.

The SMP signal represented the summed pressure output from 6 sensors placed on the thigh, in close proximity to the SEMG electrodes. No attempt was made to isolate individual quadriceps segments or account for remote muscle activities on the thigh. SMP is likely a composite of forces generated both external to and within the thigh musculature; its timing corresponded to peaks of knee extensor moment as well as ground reaction forces that compressively loaded the lower extremity during stance phase [19]. The placement of the SMP force sensors was deliberate for the present experiment, in order to measure muscle activity similarly to the EMG sensors, which required very specific locations. However, by virtue of SMP’s measurement of whole-compartment muscle activity, this precision is unnecessary for applications involving only SMP records. Changes in the compressive stiffness of muscle tissue due to contraction is manifested and distributed across the investing fascia of a limb compartment reducing the need for exacting anatomic placement of SMP sensors. For example, previous work from our group has made use of SMP in upper extremity rehabilitative devices that were donned with an even distribution of sensors about the limb [11].

Neither motion artifact nor crosstalk from the relatively low hamstring activation at these modestly stressful gait activities were likely to produce the prolonged signals seen in Fig. 3. It is possible that the foam backing might have distributed the pressure from one muscle belly to other sensors on the limb. This would further the low-pass spatial filtering of SMP records. For this reason, analysis herein is performed on a by-compartment basis.

Part of the prolonged pressure signal no doubt reflects force generation from components other than the contractile component, including fascicles and tendons within the quadriceps, whose force persists long after major activation [20]. For instance, the electromechanical delay between cessation of SEMG and cessation of torque after voluntary relaxation of the quadriceps from dynamic torque production, ranges from 249 to 300 ms [21]. The lags seen in SMP were well outside this relaxation range for all speeds except the fastest (Table 1), indicating the possibility of additional factors. An important factor contributing to lag could have been the passive tension produced in the quadriceps by flexion of the knee during mid-to-terminal stance [5]. This tension transfers a large positive power to the hip, thereby reducing the need for active propulsive energy.

4.2. Comparison with SEMG

SEMG is the most accurate method for the non-invasive registration and timing of muscle activity, and is widely used for assessing gait [1,22]. SEMG can readily delineate the sequences of supra-threshold muscle activations during gait, but its precision is dependent upon relatively non-standardized processing procedures as well as precise positioning of electrodes [23,24]. For example, onset timing of the onset during gait has been shown to be highly dependent on SEMG electrode position [24]. SMP outputs, in contrast, are low frequency signals that are relatively unaffected by variations in placement on the muscle, and their signals require much less processing than SEMG.

From surface electrodes, SEMG detects motor unit electrical impulses whose amplitude varies according to distance from the recording electrodes, and whose variability reflects the stochastic nature of the neuromotor system. As a result, the SEMG can vary widely from one stride to the next. Variance ratios of SEMGs recorded herein at various speeds averaged from 0.29 to 0.97 (Fig. 3), indicating moderate to very high variability. These results are comparable to previous results on ambulatory single muscle SEMG waveforms of 0.2–0.6 [16,25,26], determined at moderate speeds. The average VR of 0.97, obtained at the slowest speed (0.4 m/s) indicates almost complete randomness of SEMG signals from stride to stride. This inherent noisiness of SEMG may underlie the wide variance of signals recorded within and between subjects performing similar, repetitive actions [27]. The very slow gait speed elicited relatively small EMG amplitudes in comparison to background electrical signals. The high VR at 0.4 m/s may be biased by the signal to noise ratio at an exceedingly slow walking speed; a speed that was rarely utilized by our otherwise healthy subjects. Had we tested subjects that frequently engaged in this extremely...
slow gait speed due to an illness or disability, our data might have yielded a more coherent and less variable EMG pattern.

That SMP registers the active tension produced by electrical activation can be seen in the isometric trial of Fig. 2. Note that SMP rises in near synchrony with the SEMG after a lag of approximately 35 ms. The delay and waveform similarity at onset were independent of filter type, arguing against the role of filtering artifacts in the delay, which is probably electromechanical in nature. The Butterworth and moving average filters represent widely different signal processing approaches that are commonly applied to SEMG. During the several seconds of constant isometric torque, the SMP remained relatively stable, whereas the SEMG oscillated significantly. Higher noise was expected in the SEMG due stochastic motor unit recruitment, ongoing even during constant muscle force, whereas the SMP correlates with muscle force output that behaves as a low-pass system.

VRS of SMP records ranged from 0.02 to 0.1, or 6 to 30 times lower than those of SEMG. These results are expected, since SMP, being a reflection of muscle force, is a low-pass filtered version of the electrical signals [28]. Since gait strides are among the most repeatable of human motions [29], it is not surprising that the forces applied to the joints are also highly repeatable, in spite of their highly variable command signals.

Timing of quadriceps onset relative to heel strike by SMP followed that of SEMG, with similar variability, after delays of 3–12.5% of cycle. The SEMG–SMP delays generally increased with speed, and corresponded to delays ranging from 46 to 95 ms. Onset timing measured by the two modalities revealed a high degree of concordance, as indicated by ICC values ranging from 0.7 to 0.83.

The nearly perfect correlation between SEMG and SMP cession times at all but the fastest speed (ICC = 0.97–0.99, Table 1) is difficult to explain assuming that SMP cessation represents mechanical relaxation from forces that are unrelated to the initial quadriceps activation, ending 200–500 ms earlier. These forces are presumed to arise from joint reactions and possibly inertia, which may increase muscle pressure due to stretching. At least two explanations for this relationship are possible: (1) the late SMP signal, and its relaxation, was partially in response to the ‘minor’ quadriceps for this relationship are possible: (1) the late SMP signal, and its relaxation, was partially in response to the ‘minor’ quadriceps activations of the antagonist muscles. Physical Therapy in Sport 2006;7(3):122–7.

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4.3. Clinical relevance

SMP has been preliminarily validated as a reliable indicator muscle onset and cessation of quadriceps during gait, and thus may be a useful substitute or adjunct for SEMG in clinical applications that use temporal landmarks of muscle activity. Due to its relative constancy from movement-to-movement, SMP may serve as a highly reliable control signal for robotic assistive devices, for prostheses and orthoses, and for delivering biofeedback.

Presently, there are promising modeling approaches to estimating the forces produced by skeletal muscles [28], yet a simple, direct, non-invasive estimate has been unobtainable. Knowledge of the forces produced by skeletal muscles would provide clinicians with a powerful tool to supplement diagnosis, rehabilitation and outcome assessment. For example, muscular dys-coordination during gait due to neurological impairments (e.g., people with cerebral palsy, Parkinson’s, or stroke) might readily be identified while suggesting specific targets for treatment. The specific diagnosis of anterior knee pain based on SEMG analysis of the quadriceps [32], could potentially be done using the simpler modality of SMP. With further development, SMP methods may facilitate studies not only of temporal properties of muscle activation, but also of both the active and passive forces produced at working joints, which would substantially increase our understanding of muscular function.

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Conflict of interest

The authors declare no conflict.

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