Quantitative Intramuscular Myoelectric Activity of Lumbar Portions of Psoas and the Abdominal Wall During Cycling

Daniel Juker, Stuart McGill, and Peter Kropf

The purpose of this study was to quantify activation using intramuscular EMG from lumbar psoas and the three layers of the abdominal wall during several styles of cycling: normal posture (slightly flexed), upright posture, racing in flexed posture, standing up from the saddle, and standing during maximal sprint effort. Lumbar erector spinae and rectus femoris were also monitored with surface electrodes. Results demonstrated that the activity patterns were influenced by the style of cycling. Furthermore, psoas activity peaked at 14% of MVC (or less) during the upstroke phase of normal cycling but became much more active at TDC during flexed cycling (approximately 30% MVC) and approached 60% of MVC during sprinting. Generally, the abdominal wall was activated to relatively low but continual levels except during standing and sprinting. Erector spinae activity was very low at less than 5% MVC throughout the cycle until standing or sprinting styles were adopted. These normalized and scaled data on deep muscle activity during ergometer cycling provide insight into the functioning of these muscles; this information can be used to prescribe rehabilitation and training programs and can help biomechanists understand muscle activity associated with cycling.

Key Words: electromyography, ergometer, abdominal muscles

Cycling is a popular method in rehabilitation programs for knee injuries (McLeod & Blackburn, 1980), cardiac disease (Fletcher & Cantwell, 1974), and low back disorders (e.g., Schonstrom, 1988). Schonstrom (1988) suggested that bicycling is very effective for those with spinal stenosis, while Nutter (1988) summarized the apparent benefits of aerobic exercise for the low back patient. Many researchers studying cycling mechanics have collected electromyographic data using surface electrodes from trunk and leg muscles in an attempt to quantify recruitment patterns (summarized by Jorge & Hull, 1986). However, there appears to be a void in the knowledge base regarding the activity of the deep muscles of the torso and hip (specifically psoas and the abdominal wall), leaving clinicians and scientists to speculate as to their activation amplitudes and...
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Gregor, Broker, and Ryan (1991) reviewed some available cycling studies of lower extremity muscle EMG and noted that discrepancies in activation patterns have resulted from technical variations such as surface electrode placement and between/within-rider differences such as subject age and general fitness, seat height, crank cadence, workload, and rider experience. However, generally, the EMG signals reported in the cycling literature are described as either “on” or “off” and not as normalized time histories, restricting interpretation to discussion of “patterns” rather than actual levels of normalized activation. Furthermore, Raasch, Zajac, Ma, and Levine (1997) noted that during cycling, important muscles such as the iliopsoas are difficult to study (due to inaccessibility from the surface), and they proceeded to predict activation patterns using optimization. Direct measurement of activity from these muscles inaccessible from the surface would be helpful.

The purpose of this study was to quantify activation using intramuscular EMG from lumbar psoas and the three layers of the abdominal wall during several styles of cycling: normal posterior (slightly flexed), upright posture, racing in flexed posture, standing up from the saddle, and standing during maximal sprint effort. This was an invasive study and part of a much longer project that was performed for other reasons. The intent of the program was to calibrate and normalize the EMG signals to 100% maximal voluntary contraction (MVC) to facilitate interpretation of muscle forces, joint coactivation, and activation for application to athletic and clinical issues. The calibrated EMG signals collected intramuscularly during the five cycling tasks are reported here. Surface EMG data from rectus femoris and lumbar erector spinae were also recorded to enable comparison with other studies.

Methods

Data Collection

Three men (mean age 26.3 years, SD 1.5; mean height 1.72 m, SD 0.5; mean weight 66.7 kg, SD 4.5) and three women (mean age 23.7 years, SD 2.3; mean height 1.69 m, SD 0.3; mean weight 60.3 kg, SD 6.0) were recruited from a student population of elite athletes from a variety of disciplines including soccer, ice hockey, and biathlon. All subjects were fit and healthy and had never suffered from disabling low back pain. All used bicycles for transportation and most used bicycles in some component of their training. The experimental session lasted approximately 3 hr to ensure that muscle fatigue was not a confounding variable. The electrodes were in the body for approximately 4 hr. The experimental protocol was approved by the ethics committee of the Faculty of Medicine, University of Berne.

Two sets of tasks were performed in this study: The first set consisted of a series of maximal-effort isometric exertions intended to produce the largest possible amplitudes of myoelectric activity to provide a basis for EMG normalization. Subjects were given sufficient practice in performing the normalizing contractions and were provided with feedback in the form of myoelectric signals displayed in real time on a computer monitor. Real time monitoring also provided an instant subjective check on data quality. Since there was no precedent for normalizing activity of psoas, pilot work was performed to find the method to maximally activate psoas. As summarized in other work (Juker, McGill, Kropf, & Steffen, 1998), the psoas is recruited to flex the hip and appears to be unrelated to torque demands of the lumbar spine. After pilot work the following protocols were
adopted. Maximum activation of psoas (left side) was obtained by asking subjects to stand on the right leg and raise the left knee to approximately 90°, where both hands pushed down on the thigh while both maximal-effort hip flexion and spine lateral bending efforts were performed. The rectus femoris was maximally activated with subjects seated on a test bench while they performed simultaneous hip flexion and knee extension against resistance. The spine extenders were normalized during an exertion where subjects lay prone on a bench with the upper body unsupported out over the end of the table (after McGill, 1991). The feet were restrained with a tight-fitting Velcro strap. While in this position, subjects started with a slightly flexed lumbar region and then slowly extended the lumbar spine against a matched resistance on the upper back provided from an assistant. Maximal abdominal activation was obtained with the subjects starting in a bent knee sit-up posture (knees at 90°) with the feet restrained by a strap. Hands were placed on the opposite shoulder while an assistant provided a matched resistance to the shoulders (after McGill, 1991).

Three trials of each normalization exertion were performed. The instructions for the exertion were to combine maximal flexor effort with simultaneous, slow, isometric twisting efforts to either side. On occasion, some maximal activation muscle signals were obtained during other maximal exertions: For example, rectus abdominis activity may be slightly larger during the maximum psoas routine. The largest amplitude for each muscle (after signal processing) was used regardless of the activity from which it was obtained. All of these tasks will be referred to as the MVC trials.

The second set of tasks consisted of various cycling efforts on an ergometer (Monark-Trainer, Ergomed C 818 E). The tube-length of the saddle was adjusted for each subject such that the knee was fully extended with a 90° angle of the ankle during normal sitting on the saddle.

Subjects performed five different cycling tasks: For the first three tasks, the subject cycled in a slightly flexed (normal) posture, in perpendicular (upright) posture, and in racing (flexed) posture (see Figure 1). During the fourth task, the cycler stood up from the saddle with no change to the handle position (standing cycling). For all four tasks, the subjects were provided with a 90 cycle/min metronome beat although actual cycle rates were measured between 80 and 100 cycles per minute. The load was fixed at 3 kP (corresponding to 180 W). For the fifth and last task, subjects stood up from the saddle and put forth maximal sprint effort. During data collection, the EMG and synchronous video were collected for 20 cycles of each condition.

Both surface and intramuscular electrodes were used for muscles on the left side of the body. Fine wire (76 μm wire, 100 μm outer diameter gauge, purity 99.9% silver, Fa. Corradi, Milano) bipolar electrodes were made by running the wire down the outside of a 0.9 mm bore hypodermic needle where the uninsulated tips (2 mm) were inverted into the canula bore. The tips were 3 mm apart. In this way, the wires and canula were inserted into the muscle and the canula was removed, leaving the wires with the hooked end in the desired location. Wires were guided to the appropriate muscle under ultrasound images (Toshiba Tosbee SSA-24OA, 7.5 MHz probe). Two pairs of intramuscular electrodes were inserted into the psoas at the level of L3 entering the skin at approximately 10 cm from the midline and passing the lumbar fascia and quadratus lumborum muscle at an angle of approximately 45° to the sagittal plane. Subjects lay on their right sides while the psoas electrodes were inserted. Xylocaine 1% (2–3 ml) was injected to anesthetize the skin and fascia at entry. Intramuscular electrodes were also placed in external oblique, internal oblique, and transverse abdominis midway between the linea semilunaris and the midline laterally and at the transverse level of the umbilicus. Surface electrodes (Beckmann bipo-
lar, Ag Ag-C1, with a center-to-center distance of 2.5 cm) were placed on the following muscles: rectus abdominis, 3 cm lateral to the umbilicus; external oblique, approximately 15 cm lateral to the umbilicus and at the transverse level of the umbilicus; internal oblique, below the external oblique electrodes and just superior to the inguinal ligament; lumbar erector spinae, 3 cm lateral to the L3 spinous process; and rectus femoris, over the belly of the muscle approximately 8 cm below the inguinal ligament.

All raw myoelectric signals were preamplified (gain = 1000, common mode rejection ratio greater than 100 dB at 50 Hz, filtered to produce a band-width of 4 to 20,000 Hz). Signals were further filtered (10–500 Hz to minimize artifact) and amplified (2 to 8 times) with custom-made analog instrumentation to produce signals of approximately ±5 V. The sagittal plane view of the subjects was also filmed on video tape.

Data Reduction

All myoelectric signals were A/D converted (12 bit resolution) at 1024 Hz. Signals were then digitally full-wave rectified and low-pass filtered (single pass, Butterworth) at a cut-
off frequency of 3 Hz in an attempt to match the frequency response of muscle, and then were amplitude normalized to the maximum activity observed during the MVC trials. The cutoff frequency of 3 Hz was chosen in the following way: Olney and Winter (1985) reported the frequency response of the rectus femoris to be between 1.0 and 2.8 Hz during walking, whereas Milner-Brown, Stein, and Yemm (1973) reported approximately 3 Hz in the first dorsal interosseous. In addition, the 3 Hz cutoff produced an impulse response (time to peak) of 53 ms, which is comparable with the 30–90 ms contraction times reported by Buchthal and Schmalbruch (1970). In this way, the low-pass filter created the electromechanical delay between the electrical event (EMG) and the mechanical event (muscle force).

Six pedal cycles were recorded for each subject/task/trial once subjects reached steady-state cycling. Time histories of normalized muscle activation were calculated by ensemble averaging three cycles from each condition and from each subject. Time bases were normalized to 100% of the cycle, beginning and ending with the left crank at top-dead-center (TDC). While electrode reliability was a concern in some other dynamic activities (other than cycling), signals appeared to be stable during the cycling tasks; this is a limitation with indwelling electrodes.

**Results**

Intramuscular normalized muscle activity records are shown in Figure 2 for the abdominal wall (oblique and transverse muscles) and psoas, together with signals from erector spinae and rectus femoris collected with surface electrodes. Activity patterns are influenced by the style of cycling.

![Figure 2 — Muscle activation profiles normalized to the crank cycle starting with left top center through a full revolution to left TDC.](image-url)
Psoas activity peaked at 14% of MVC (or less) during the upstroke phase of normal cycling but became much more active at TDC during flexed cycling (approximately 30% MVC) and approached 60% of MVC during sprinting. Generally the abdominal wall was activated to relatively low but continual levels (e.g., less than 8% MVC) except during standing and sprinting. Erector spinae activity was very low (from a clinical perspective).
at less than 5% MVC throughout the cycle until standing or sprinting styles were adopted. Variability is indicated in Table 1, where mean amplitude values together with their standard deviation were calculated when the left and right crank arms reached top dead center. (Note that mean values are slightly different than those shown in the graphs; these analyses were conducted at different times and the choice of which video frame corresponded
to crank TDC varied slightly.) Those signals with small mean values had correspondingly lower standard deviations. Most standard deviation values were below 10% MVC, while the largest mean amplitude (observed in psoas during sprint cycling) of 49% MVC had the largest standard deviation of 26% MVC. Even within what appeared to be steady-state
cycling, muscle activation amplitudes varied from stroke to stroke as seen in Figure 3 of a subject standing on the pedals cycling.

**Discussion**

These data contribute information on normalized activation in muscles that are not accessible by surface electromyography. From a functional perspective, the observed activity of psoas is consistent with its role as primarily a hip flexor (Juker et al., 1998; Santaguida & McGill, 1995) while the abdominal wall appears to stabilize the torso with continual low activation (Cholewicki & McGill, 1996; Richardson, Toppenberg, & Jull, 1990). Flexed postures appear to increase psoas activity at TDC of the pedal while sprinting elevates activity commensurate with the increased hip flexor demand. It appears that patients should not consciously flex the torso during cycling when the psoas, hip, or low back is of concern, for example, following injury or surgery.

There are few data available to compare to the data of this study. Jorge and Hull (1996) presented polar plots of the activity (strictly on/off) of lower extremity muscles in which they summarized rectus femoris activity. Generally, the muscle became active from approximately 90–180° from TDC to approximately 260–360°. Our data on rectus femoris (note that our electrodes were placed proximally, very close to the inguinal ligament) show that phasic activity is more characteristic of the “bottom” of the crank cycle (90°) in normal postures, while flexed postures tend to delay the phasic activity to later in the recovery phase of the cycle, consistent with Jorge and Hull’s (1986) summary. Generally, rectus femoris precedes psoas activity in the cycle, which again makes qualitative “sense” given that the rectus femoris has both hip flexor and knee extensor responsibilities while the psoas is responsible for hip flexion (McGill, Juker, & Kropf, 1996). It is also interesting to observe that the predictions of psoas activity by Raasch et al. (1997), based on optimization algorithms, are similar in pattern to our upright cy-

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**Table 1  Muscle Activation Normalized to 100% MVC (Mean and Standard Deviation)**

<table>
<thead>
<tr>
<th></th>
<th>Psoas 1</th>
<th>Psoas 2</th>
<th>Rect fem</th>
<th>Erector sp</th>
<th>Rect ab</th>
<th>Ext obl.</th>
<th>Int obl.</th>
<th>Transverse abdom.</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Left crank top</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Normal</td>
<td>4(5)</td>
<td>1(1)</td>
<td>7(4)</td>
<td>4(3)</td>
<td>10(7)</td>
<td>4(5)</td>
<td>2(2)</td>
<td>4(3)</td>
</tr>
<tr>
<td>Upright</td>
<td>6(4)</td>
<td>2(1)</td>
<td>7(3)</td>
<td>3(2)</td>
<td>10(7)</td>
<td>3(3)</td>
<td>2(2)</td>
<td>4(3)</td>
</tr>
<tr>
<td>Racing flexed</td>
<td>29(15)</td>
<td>14(7)</td>
<td>12(6)</td>
<td>3(3)</td>
<td>11(6)</td>
<td>6(3)</td>
<td>4(3)</td>
<td>5(5)</td>
</tr>
<tr>
<td>Standing</td>
<td>6(7)</td>
<td>1(1)</td>
<td>12(6)</td>
<td>8(4)</td>
<td>5(1)</td>
<td>3(1)</td>
<td>10(15)</td>
<td>6(1)</td>
</tr>
<tr>
<td>Sprint</td>
<td>49(26)</td>
<td>14(13)</td>
<td>14(10)</td>
<td>32(15)</td>
<td>10(5)</td>
<td>28(16)</td>
<td>8(9)</td>
<td>9(9)</td>
</tr>
<tr>
<td><strong>Right crank top</strong></td>
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<td></td>
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<td></td>
<td></td>
</tr>
<tr>
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<td>6(7)</td>
<td>15(20)</td>
<td>4(3)</td>
<td>10(6)</td>
<td>4(2)</td>
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</tr>
<tr>
<td>Upright</td>
<td>5(5)</td>
<td>3(1)</td>
<td>11(15)</td>
<td>4(2)</td>
<td>10(7)</td>
<td>6(5)</td>
<td>3(2)</td>
<td>6(2)</td>
</tr>
<tr>
<td>Racing flexed</td>
<td>10(8)</td>
<td>3(3)</td>
<td>6(3)</td>
<td>2(1)</td>
<td>10(7)</td>
<td>5(5)</td>
<td>2(2)</td>
<td>5(3)</td>
</tr>
<tr>
<td>Standing</td>
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<td>1(1)</td>
<td>11(10)</td>
<td>11(9)</td>
<td>6(2)</td>
<td>8(6)</td>
<td>7(6)</td>
<td>7(1)</td>
</tr>
<tr>
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<td>3(2)</td>
<td>14(9)</td>
<td>20(17)</td>
<td>10(6)</td>
<td>12(13)</td>
<td>10(13)</td>
<td>14(8)</td>
</tr>
</tbody>
</table>

*Note.* Rectus femoris, erector spinae, and rectus abdominis were surface electrodes.
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Figure 3 — Stroke-to-stroke variability in muscle activation levels seen from five cycles of one subject standing on the pedals.

This study has several limitations. This was an invasive study, which reduced subject numbers. Subjects were young and very fit, restricting the relevance of these data to older or injured individuals. Furthermore, all of the usual limitations of using fine wire electrodes pertain to this data, such as monitoring only a small portion of the entire muscle and the possibility of wire migration during contraction. In the case of psoas, we made two insertions, with one electrode pair systematically lower than the other. The second channel was in another layer of the muscle, although this was not quantified. Major wire migrations sufficient to change the raw signal were evident during some activities (but not the cycling activities reported in this work) that necessitated recalibration using the calibration tasks described in the Methods section. Finally, internal electrodes have also been demonstrated to affect gross movement, at least in children during walking (Young, Rose, Biden, Wyatt, & Sutherland, 1989). For this reason, together with hardware limitations, we limited the total number of electrodes on each subject.

The normalized and scaled data on deep muscle activity during ergometer cycling presented here will assist clinicians, fitness professionals, and biomechanists in understanding the mechanics associated with cycling activity.

References


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Acknowledgments

We thank the subjects who participated in this study, Professor H. Hoppeler and Dr. T. Steffen for scientific support, and Maria Keiser and Michael Kientsch for technical assistance. We gratefully acknowledge the financial support of the Federal Sports School and the Federal Committee of Sports, Switzerland, and the Natural Sciences and Engineering Research Council, Canada.