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Pelvic floor muscle electromyography during different running speeds: an exploratory and reliability study

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Abstract
Purpose Stress urinary incontinence (SUI) affects women of all ages including young athletes, especially those involved in high-impact sports. To date, hardly any studies are available testing pelvic floor muscles (PFM) during sports activities. The aim of this study was the description and reliability test of six PFM electromyography (EMG) variables during three different running speeds. The secondary objective was to evaluate whether there was a speed-dependent difference between the PFM activity variables.
Methods This trial was designed as an exploratory and reliability study including ten young healthy female subjects to characterize PFM pre-activity and reflex activity during running at 7, 9 and 11 km/h. Six variables for each running speed, averaged over ten steps per subject, were presented descriptively, tested regarding their reliability (Friedman, ICC, SEM, MD) and speed difference (Friedman).
Results PFM EMG variables varied between 67.6 and 106.1 %EMG, showed no systematic error and were low for SEM and MD using the single value model. Applying the average model over ten steps, ICC (3,*k*) were >0.75 and SEM and MD about 50 % lower than for the single value model. Activity was found to be highest in 11 km/h.
Conclusion EMG variables showed excellent ICC and very low SEM and MD. Further studies should investigate inter-session reliability and PFM reactivity patterns of SUI patients using the average over ten steps for each variable as it showed very high ICC and very low SEM and MD. Subsequently, longer running distances and other high-impact sports disciplines could be studied.

Keywords Jogging · Pelvic floor · Reproducibility · Sports · Stress urinary incontinence

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Introduction

Stress or "activity-related" urinary incontinence means the complaint of involuntary loss of urine due to effort or physical exertion, e.g., sporting activities, or when sneezing or coughing [1] and affects women of all ages [2]. Sporting activities, which involve high impact, result in the highest prevalence of stress urinary incontinence (SUI) and also affect young athletes [3]. Goldstick and Constantini [4] state in their review that top female athletes report a high prevalence of urinary incontinence, especially during sports but also during daily activities, and that the prevalence of urinary incontinence ranges from 28 to 80 %, with the highest prevalence in high-impact sportswomen such as trampolinists, gymnasts, aerobic gymnasts, hockey players and ballet dancers. Women who attend gym and perform high-impact exercise have a greater prevalence of urine loss than women who do not perform any high-impact exercises [5].

Activities, which typically provoke incontinence, raise the intra-abdominal pressure and the impact loading on the pelvic floor muscles (PFM) [5]. High-impact physical activities, where both feet are off the ground at the same time (e.g., when jumping or running), involve abrupt

repeated increase in abdominal pressure [5]. To date, few studies are available investigating PFM activity during high-impact loads and hardly any of those concern functional whole-body movement situations, e.g., sports activities. Luginbuehl et al. [6] found PFM electromyography (EMG) pre-activity of 72.1 %EMG [EMG normalized to maximal voluntary contraction (MVC)] at 50 ms prior to the heel strike during running at 8 km/h, which means a PFM activity of approximately 40 %EMG higher than PFM activity during standing without any voluntary contraction. They also found an immediate strong increase up to a mean maximal PFM EMG activity of 124.3 %EMG within 214.2 (±51.8) milliseconds (ms) after the heel strike, which suggests involuntary and therefore PFM reflex activity during running impact loads [6]. Further studies also found PFM pre-activity and reactivity [7, 8], however, those PFM activity measurements all took place in static positions of the subjects (standing or supine).

PFM reflex activity might concern stretch reflexes, i.e., a stretch–shortening cycle muscle function. According to Komi [9], the stretch–shortening cycle proceeds in three phases, namely pre-activity, eccentric lengthening and concentric contraction. Because of the eccentric lengthening a reactive and stronger contraction can follow, which allows the muscle to generate more strength in shorter time [9, 10]. Stretch reflexes can be classified according to their latencies—i.e., reflex peaks—and are characterized by slow, mid and long latency responses and long latency succeeding responses in relation to an impact, e.g., the initial ground contact during running [10].

More knowledge of PFM function is essential to get a better understanding of the pathophysiology of SUI and as a result to develop more precise diagnostic methods regarding PFM activity and contraction components. First and foremost PFM function has to be clarified for functional movements with short impacts typically provoking SUI such as running or jumping [1], and not only for non-functional isolated test situations such as MVC in supine [11, 12].

The aim of the present study was to investigate and describe PFM activity during high-impact sports activities under various conditions. The specific goals were the reliability test of six PFM EMG variables during three different running speeds. The secondary objective was to evaluate whether there was a speed-dependent difference between the PFM EMG variables.

Materials and methods

Study design

This trial was designed as an exploratory and intra-session retest reliability study to characterize PFM activity during running at three different speeds. It focuses on the description and reliability of six previously defined EMG variables of pre-activity and reflex activity and, as a second outcome, on the difference between the three speeds regarding those variables.

The study was conducted in accordance with the Declaration of Helsinki, and all subjects gave written informed consent. Following an agreement with the ethics committee, approval was not required as the investigation concerned a physiotherapy-relevant reliability low-risk study.

Subjects

Ten female subjects were recruited from the Bern University of Applied Sciences and were included on condition that they were aged between 20 and 35 years, were nulliparous and anamnestically healthy, had a BMI between 20 and 30 kg/m^2, were physically able to cope with the requirements of the testing procedure and were experienced and familiar with treadmill running. Subjects, who had their period or who had had surgery in the urogenital region as well as those with acute vaginal infection, incontinence, pelvic floor complaints, pain during running, acute back or joint pain, acute injury of the lower extremity, or nickel or latex allergy were excluded. All subjects were trained in MVC of their PFM as the learning of a correct isolated (maximal) PFM contraction was part of the practical program of PFM rehabilitation in their professional physiotherapy or midwifery education.

Instrumentation

The treadmill was a Kettler Marathon TX1 device (Ense-Parsit, Germany). All subjects had to perform their running at the speeds of 7, 9 and 11 km/h and 1° inclination. A vaginal surface EMG probe (Periform®, Neen, UK-Oldham Lancashire) was used to measure PFM activity. The single reference adhesive surface electrode (Ambu Blue Sensor N, Ballerup, Denmark) was fixed on the right iliac crest according to the SENIAM recommendations [13]. A force-sensitive resistor footswitch (2-FSR, Noraxon European Service Center, Cologne, Germany) was used to identify the initial contact (T0), i.e., the initial time point of the impact and beginning loading phase and strain of the PFM. The footswitch consists of two FSR sensors, which were fixed with adhesive tape on the right heel and ball of the big toe to optimally capture the initial contact. Electrodes and footswitches were connected to the transmitter by short wire, which was fixed at the back of the subjects. The signals were sent wirelessly to the receiver (TeleMyo 2400 G2, Noraxon European Service Center, Cologne, Germany).

Procedures

Demographics (age, weight, height and body mass index) were determined and after emptying their bladder the subjects were equipped with the footswitch and EMG reference electrode. The subjects were instructed in the vaginal insertion of the surface EMG probe using ultrasound lubrication and then performed the insertion themselves. The subjects wore a loose running suit and were barefoot, as the various shock absorption systems of the subjects' individual running footwear could influence running ground reaction forces and force transmission [6].

PFM EMG was measured twice for 15 s without any voluntary contraction and twice for 5 s during MVC (contraction maximal as possible) in a standing position. Between the single measurements, a 15-s break was taken. The MVC testing while standing was chosen instead of the usual MVC testing in a supine position [6, 14] because it seemed more functional in comparison with running. Thereafter, the subjects performed a warm-up of walking (5 km/h) for 30 s, then running at 7, 9 and 11 km/h consecutively until they reached a steady state. As soon as they reached the steady state at the respective speed, the data acquisition was started: EMG and footswitch signals were measured continuously for 15 s and the first 10 step cycles of the right leg were analyzed. The subjects were instructed to run, to breathe as normally as possible, not to activate their PFM voluntarily and not to talk during the measurements. Between the measurements of the different speeds, the treadmill was stopped, followed by a 1-min break until restarting the same procedure with the next speed.

Data reduction

EMG and footswitch signals were sampled at a rate of 2 kHz [sampling interval (dt) equals 0.5 ms] using a 12-bit analog-to-digital converter (ME-2600i, SisNova Engineering, Zug, Switzerland) and the software package "Analoge und digitale Signalverarbeitung" (ADS) version 1.12 (uk-labs, Kempen, Germany).

The EMG signals were initially first-order high-pass filtered with a cutoff frequency of 10 Hz by EMG preamplifier leads to reject or eliminate artifacts and later digitally low-pass filtered by ADS software with a cutoff frequency of 1 kHz (second-order zero-lag Butterworth filter, 24 dB/octave filter steepness) to avoid aliasing. Second, to identify amplitude peaks during MVC, EMG was calculated as RMS (200 ms moving window). 100 % of EMG equals the average of the two peak amplitude values during the two 5-s sessions of MVC. Third, EMG variables were calculated as RMS values within each 30-ms interval [10, 15, 16], averaged over 10 steps and normalized to peak MVC (%EMG). The different activity variables are described in Table 1. All variables were analyzed using the software package ADS.

PFM EMG data during standing without any voluntary contraction and MVC were averaged over test and retest, and PFM EMG data of running over ten steps for each subject.

To determine PFM EMG activity during the initial phase of ground contact, the approach according to Fleischmann et al. [10, 15] was chosen: As basically no clear and reproducible reflex peaks could be determined visually on the rectified EMG of consecutive steps, mean amplitudes for fixed 30-ms intervals covering the phase in which reflex activity is expected to occur were calculated. Therefore, reflex phase amplitudes were calculated between 30 and 60 ms (short latency response), 60–90 ms (mid latency response), 90–120 ms (long latency response) and 120–150 ms (long latency succeeding response). Additionally, pre-activity was computed during the interval between −30 ms and T0 [10, 15].

Following the study protocol of Fleischmann et al. [15], who calculated EMG amplitudes of shank muscles between touchdown to 150 ms of ground contact in 30-ms time windows for lateral jumps from four different distances, the same 30-ms time intervals were calculated for all three running speeds in the present study.

Ten strides of under extremity muscles' EMG data provide a very high level of stability of a given subject relative to the variability across subjects [17]. Therefore, the analysis of EMG data of 10 steps was also chosen for this study.

Statistical analysis

A total sample size of $N \geq 9$ and an associated actual power of 0.83 were computed as a bivariate normal model by means of G*Power software [18] based on the following assumptions: one-tailed test, correlation coefficient of alternative hypothesis: 0.75; alpha error probability: 0.05; power (1 − beta error probability): 0.80; correlation coefficient of null hypothesis: 0.00.

Descriptive statistics were performed for each variable [mean, standard deviation (SD)].

The reliability test followed the three-step suggestions of Weir [19]: To identify possible systematic errors between the repeated measures, the Friedman test for n-dependent samples to compare EMG variables over the ten steps was applied. Reliability was calculated for single measures (absolute agreement) and average measures (consistency) with the two-way random intraclass correlation coefficients [ICC (3,1) and (3,k)] (i.e., relative reliability) which do not consider systematic error. The absolute standard error of measurement ($SEM = SD \times \sqrt[2]{1 - ICC}$; i.e., absolute reliability), the relative SEM related to the mean (SEM%),

Table 1 Labels, units, and description of activity and time variables derived from electromyography (EMG) and footswitch

Variable	Unit	Description	To identify
T−30–0	%EMG	Mean EMG activity between T0 and minus 30 ms	The mean pre-activity between T0 and minus 30 ms as a regulatory component of anticipation
T0–30	%EMG	Mean EMG activity between T0 and 30 ms	The mean EMG amplitude between the initial contact and 30 ms as the initial ground-contact phase, the interval preceding latency responses
T30–60	%EMG	Mean EMG activity between 30 ms and 60 ms after T0	The mean EMG amplitude between 30 and 60 ms to detect short latency response (SLR) as a characterization of reflex activity during a stretch–shortening cycle
T60–90	%EMG	Mean EMG activity between 60 ms and 90 ms after T0	The mean EMG amplitude between 60 and 90 ms to detect mid latency response (MLR) as a characterization of reflex activity during a stretch–shortening cycle
T90–120	%EMG	Mean EMG activity between 90 ms and 120 ms after T0	The mean EMG amplitude between 90 and 120 ms to detect long latency response (LLR) as a characterization of reflex activity during a stretch–shortening cycle
T120–150	%EMG	Mean EMG activity between 120 ms and 150 ms after T0	The mean EMG amplitude between 120 and 150 ms to detect long latency succeeding response (LLR2) as a characterization of reflex activity during a stretch–shortening cycle

additionally the absolute minimal difference ($MD = SEM \times 1.96 \times \sqrt[2]{2}$) needed to be considered real, and relative MD related to the mean (MD%) were computed. SEM and MD were calculated twice, once related to the ICC (3,1) and once to the ICC (3,*k*).

For the evaluation of relative reliability ICC values benchmarks presented by Shrout and Fleiss [20] were used: Above 0.75 represents excellent, 0.40–0.75 represents fair to good and below 0.40 represents poor reliability.

As to the secondary outcome concerning the differences of the EMG variables between and within the three running speeds, analyses of variance (Friedman test and post hoc Wilcoxon test) were performed.

The level for significances was set to $P \leq 0.05$ (Bonferroni correction $P < 0.017$ and 0.003). All statistics were calculated with IBM SPSS 20 for Windows (SPSS, Inc; Chicago, IL, USA).

Results

The ten included subjects had a mean (±SD) age of 24.9 years (±3.3), weight of 59.5 kg (±7.7), height of 1.7 m (±0.1) and body mass index of 21.6 kg/m^2 (±2.7).

Descriptive statistics and reliability calculations of the six EMG variables of the three different running speeds, 7, 9 and 11 km/h are presented in Table 2 and Fig. 1.

PFM EMG during standing without any voluntary contraction showed a mean of 29.6 %EMG. The values of the PFM EMG variables regarding the rather slow running speeds of 7 and 9 km/h are similar and lie between 67.6 and 88.4 %EMG. Only running at 11 km/h leads to higher PFM values rising up to 106.1 %EMG. The means of the PFM EMG variables increase from pre-activity (75.4–91.6 %EMG) to T30–60 (84.9–106.1 %EMG) and then decrease to T120–150 (67.6–70.3 %EMG). This increase and decrease was significant for 11 km/h only ($P < 0.001$).

The analysis of systematic errors within repeated measures by the Friedman test revealed only non-significant values for all variables.

ICC for single values (3,1) ranged from 0.24 to 0.56 at a speed of 7 km/h, 0.09 to 0.57 at 9 km/h and 0.25 to 0.62 at 11 km/h. SEM (SEM%) based on this ICC (3,1) are generally low and range between 3.5 (4.3) and 4.9 (6.9) %EMG for 7 km/h, 2.7 (3.5) and 5.6 (6.8) %EMG for 9 km/h and 4.6 (5.2) and 6.7 (6.5) %EMG for 11 km/h.

In contrast ICC for averaged values (3,*k*) ranged from 0.76 to 0.93 at a speed of 7 km/h, 0.49 to 0.93 at 9 km/h and 0.77 to 0.94 at 11 km/h. SEM (SEM%) based on this ICC (3,*k*) are generally low and range between 1.9 (2.4) and 2.5 (2.9) %EMG for 7 km/h, 1.9 (2.6) and 2.4 (3.2) %EMG for 9 km/h and 2.1 (2.5) and 2.9 (4.1) %EMG for 11 km/h.

Accordingly to SEM (SEM%), the MD (MD%) related to ICC (3,1) were generally higher than related to ICC (3,*k*). MD (MD%) related to ICC (3,1) ranged from 9.8 (12.0) to 13.5 (19.2) %EMG for 7 km/h, 7.4 (9.8) to 15.5 (18.9) %EMG for 9 km/h and 12.7 (14.4) to 18.6 (18.1) %EMG for 11 km/h. MD (MD%) related to ICC (3,*k*) ranged from 5.2 (6.6) to 7.0 (7.9) %EMG for 7 km/h, 5.4 (7.1) to 6.6 (9.0) %EMG for 9 km/h and 5.9 (6.8) to 8.0 (11.4) %EMG for 11 km/h.

As to the secondary outcome significant differences ($P < 0.05$) were presented for T−30–T0, T0–30, and T30–60, only. In detail, there are no differences for these variables between 7 and 9 km/h, however, between 7 and 11 and 9 and 11 km/h (Fig. 1).

Table 2 Descriptive statistics (mean ± SD), reliability indexes [ICC (3,1); ICC (3,*k*)], ICC-related SEM, SEM%, MD, MD%; and test for systematic error (Friedman) for activity variables derived from pelvic floor muscles' electromyography during running at 7, 9 and 11 km/h

Variable	Mean, %EMG	SD, %EMG	ICC 3,1	ICC 3,*k*	SEM, ICC 3,1 %EMG	SEM%, ICC 3,1	SEM, ICC 3,*k* %EMG	SEM%, ICC 3,*k*	MD, ICC 3,1%EMG	MD%, ICC 3,1	MD, ICC 3,*k* %EMG	MD%, ICC 3,*k*	Friedman, *P* value
7 km/h													
T−30–0	76.1	4.3	0.27	0.79	3.7	4.8	2.0	2.6	10.2	13.4	5.5	7.2	0.097
T0–30	81.8	4.1	0.25	0.77	3.5	4.3	2.0	2.4	9.8	12.0	5.4	6.6	0.598
T30–60	88.4	5.2	0.24	0.76	4.5	5.1	2.5	2.9	12.5	14.2	7.0	7.9	0.115
T60–90	80.8	6.0	0.40	0.87	4.6	5.7	2.2	2.7	12.9	15.9	6.0	7.4	0.952
T90–120	73.2	5.0	0.38	0.86	3.9	5.4	1.9	2.5	10.9	14.9	5.2	7.1	0.687
T120–150	70.4	7.3	0.56	0.93	4.9	6.9	2.0	2.8	13.5	19.2	5.5	7.8	0.253
9 km/h													
T−30–0	75.4	2.8	0.09	0.49	2.7	3.5	2.0	2.7	7.4	9.8	5.5	7.4	0.709
T0–30	81.1	4.8	0.30	0.81	4.0	4.9	2.1	2.6	11.1	13.7	5.8	7.1	0.101
T30–60	84.9	8.3	0.55	0.92	5.6	6.6	2.3	2.7	15.5	18.3	6.4	7.5	0.091
T60–90	80.3	8.2	0.57	0.93	5.4	6.7	2.2	2.7	14.9	18.6	6.0	7.5	0.368
T90–120	74.0	6.7	0.43	0.87	5.1	6.8	2.4	3.2	14.0	18.9	6.6	9.0	0.442
T120–150	67.6	6.8	0.55	0.92	4.5	6.7	1.9	2.9	12.6	18.6	5.4	7.9	0.312
11 km/h													
T−30–0	91.6	5.5	0.25	0.77	4.8	5.2	2.7	2.9	13.2	14.4	7.4	8.0	0.052
T0–30	97.8	6.8	0.41	0.87	5.2	5.3	2.4	2.5	14.5	14.8	6.7	6.8	0.086
T30–60	106.1	10.8	0.62	0.94	6.7	6.3	2.6	2.5	18.6	17.5	7.3	6.9	0.359
T60–90	86.1	6.6	0.46	0.90	4.8	5.6	2.1	2.5	13.4	15.6	5.9	6.9	0.670
T90–120	80.3	7.0	0.48	0.90	5.1	6.3	2.2	2.7	14.0	17.5	6.1	7.6	0.054
T120–150	70.3	6.6	0.52	0.81	4.6	6.5	2.9	4.1	12.7	18.1	8.0	11.4	0.082

Mean arithmetic mean, *SD* standard deviation, *ICC* intraclass correlation coefficient, *SEM* absolute standard error of measurement, *SEM%* relative standard error of measurement, *MD* absolute minimal difference, *MD%* relative minimal difference, *Friedman* Friedman test non-parametric repeated measures comparison

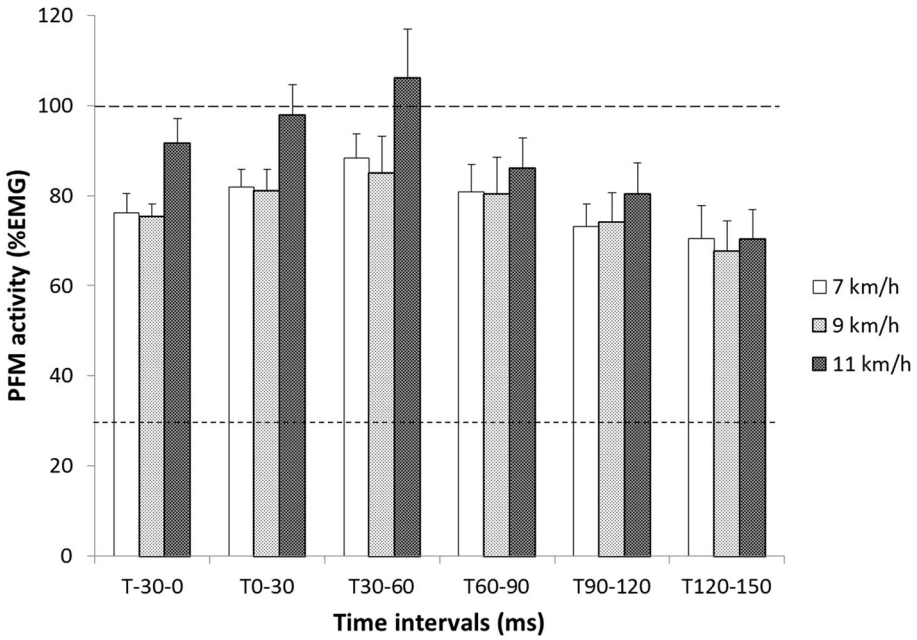

Fig. 1 Means and standard deviations of PFM activity variables (time intervals of 30 ms) of three running speeds in %EMG

Discussion

Primary outcome

All PFM EMG variables of all running speeds show clearly higher values than PFM activity during standing without any voluntary contraction, whose mean of 29.6 %EMG being similar to the findings of Luginbuehl et al. [6] and Lauper et al. [21]. The higher values than during standing without any voluntary contraction suggest a PFM pre-activity and reflex activity during running. There is an increase of activity from T−30–0 to T30–60, and a decrease from T30–60 to T120–150 during 11 km/h. According to the peak activity during T30–60 and the corresponding short latency response, it could be hypothesized that during the highest speed (11 km/h) a fast monosynaptic reflex [22] follows the impact of initial contact.

As the Friedman test revealed only non-significant values for all variables, a systematic error within repeated measures can be excluded. Consequently, ICC (3,1) and (3,*k*) are correctly chosen tests as they only consider random error [19]. The rather low ICC (3,1) can be considered of little importance as the SEM (SEM%) and MD (MD%) relating to the ICC (3,1) show really low values accounting for high reliability. As expected, average values of ICC (3,*k*) show higher values, and SEM (SEM%) and MD (MD%) related to ICC (3,*k*) show lower values as an average over ten steps reduces systematic error [17].

Therefore, the ICC (3,*k*) shows higher values than the ICC (3,1) and (3,*k*)-related SEM (SEM%) and MD (MD%) approximately 50 % lower values. With the exception of one, all ICC (3,*k*) are higher than 0.75 and meet the highest benchmarking excellent and the statistical power requirements of >0.8.

Grape et al. [14] showed good to high PFM EMG retest reliability regarding average activity, peak, work and baseline [ICC (2.1) = 0.83–0.96] of isolated PFM contractions for healthy nulliparae aged 20–35 years. Auchincloss and McLean [23] investigated between-trial and between-day reliability of EMG data (peak EMG amplitudes) recorded from the PFM during the functional task of coughing using two different probes. Overall, they found that between-trial reliability was fair to high for the Femiscan™ [ICC (3,1) = 0.58–0.98] and good to high for the Periform™ [ICC (3,1) = 0.80–0.98], however, between-day reliability was generally poor for both vaginal probes [ICC (3,1) = 0.08–0.84]. To the authors' knowledge, the only study investigating the reliability of PFM EMG during functional whole-body movements was conducted by their own research group: Luginbuehl et al. [6] tested eight PFM EMG variables of pre-activity and reflex activity during treadmill running at 8 km/h for reliability. Six EMG variables showed good reliability and two (regression variables) showed moderate to good ICC (3,1) values. Auchincloss and McLean [24] investigated whether vaginal probes may induce changes in PFM recruitment by the very presence of the probes and found that the Femiscan™ and Periform™ vaginal probes do not influence PFM activation amplitude during a PFM MVC task.

Secondary outcome

The higher values for the EMG variables of T−30–T0, T0–30, and T30–60 of the faster running speed of 11 km/h rising up higher than MVC related to the slower running speeds could be the response to the higher ground reaction forces and therefore higher impacts during a faster running speed [25] and suggest that the higher PFM activity compared to MVC owes to reflexive and reactive force generation during running.

Future studies should test whether higher running speeds than 11 km/h would lead to even higher PFM EMG activities, i.e., if their values would generally rise above MVC.

Limitations

Crosstalk

Although EMG is a reliable method of assessing PFM activity in healthy women [6, 14], crosstalk can confound the interpretation of EMG recordings using a bipolar surface electrode arrangement [26]. Peschers et al. [27] showed in their investigation that additional contraction of the gluteal muscles together with PFM leads to significantly higher PFM EMG compared to isolated PFM contraction. However, crosstalk for PFM during sports activities such as running has not yet been examined and therefore is difficult to estimate. As there is, among others, muscular activity in the hip adductors and gluteus maximus during running [28], crosstalk cannot be excluded and therefore should be subject to further investigations.

Keshwani and McLean [29] recommend using smaller electrodes and differential electrode configurations to decrease the likelihood of recording crosstalk. The electrodes applied in the current study employ a "faux differential" configuration [29]. However, the only commercially available intravaginal probe with differential electrode configuration is the Femiscan [29], which seems not appropriate for applying during running due to its size and shape. To minimize crosstalk, a 3-pol-STIMPON® electrode (Innocept Biobedded Medizintechnik GmbH, Gladbeck, Germany) in a differential configuration could be recommended for future studies. As it has smaller electrodes and is totally inserted into the vagina and adapts its shape individually to the vaginal cavity, this probe could be ideal to minimize crosstalk.

Motion artifacts

Movement of the probe relative to the underlying skin temporarily distorts the EMG signal and creates motion artifacts [29]. However, an implemented 10 Hz high-pass filter in the preamplifier, using well-fixed short wires between the vaginal electrode and transmitter, and wireless technology minimized motion artifacts induced by the movement of the EMG electrodes while the participants were running on the treadmill. In addition, the raw EMG data was visually controlled (e.g., baseline shifts) by an experienced researcher, who did not identify any abnormal EMG patterns. Spectrum analysis (Fast Fourier Transformation) of the EMG data did not reveal any movement or alternating current hum-related artifacts.

Barefoot running

As footwear might influence running and force transmission [30], the subjects ran barefoot. However, from a biomechanical viewpoint, barefoot running can change the landing pattern: Habitually, barefoot runners tend to use the forefoot running pattern, whereas most of the shoed runners use the heel strike pattern [31]. Forefoot strikes lead to a significant reduction in the loading rate [31]. In the present study, most subjects were not used to barefoot running and some subjects repeatedly changed their landing pattern within a measurement period. Therefore, an influence of the landing patterns and the subsequent changes in loading rates [31] and their influence on EMG measurements cannot be excluded. A future study should apply standardized running shoes or, when running barefoot, previously familiarize the subjects with a heel strike running pattern.

Conclusion

Up to now, research focused on non-functional isolated test situations such as concentric and isometric voluntary muscle action forms, which lead to lift and squeeze [11, 12] and physical therapy concentrated mainly on muscle hypertrophy training [3]. This study showed excellent intra-session ICC and very low SEM and MD of PFM EMG variables of pre-activation and reflexive function during running. Future research should focus on the inter-session reliability of the variables of the present study. A next step would be a similar investigation in women suffering from SUI to get insight in PFM reactivity patterns of the affected. For such investigations, the findings of the current study recommend to average ten steps as a very precise and reliable measure of PFM activity during running. In addition, studies with longer running distances including healthy and affected women to test repetitive strain on the PFM and therefore PFM reactive strength endurance function as well as investigations regarding other high-impact sports disciplines typically provoking SUI such as trampoline jumping, track and field or gymnastics [32] would be of high interest.

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Compliance with ethical standards

Conflict of interest None.

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