

Intra-session test–retest reliability of pelvic floor muscle electromyography during running

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Abstract

Introduction and hypothesis The prevalence of female stress urinary incontinence is high, and young adults are also affected, including athletes, especially those involved in “high-impact” sports. To date there have been almost no studies testing pelvic floor muscle (PFM) activity during dynamic functional whole body movements. The aim of this study was the description and reliability test of PFM activity and time variables during running.

Methods A prospective cross-sectional study including ten healthy female subjects was designed with the focus on the intra-session test–retest reliability of PFM activity and time variables during running derived from electromyography (EMG) and accelerometry.

Results Thirteen variables were identified based on ten steps of each subject: Six EMG variables showed good reliability (ICC 0.906–0.942) and seven time variables did not show good reliability (ICC 0.113–0.731). Time variables (e.g. time difference between heel strike and maximal acceleration of vaginal accelerator) showed low reliability. However, relevant PFM EMG variables during running (e.g., pre-activation, minimal and maximal activity) could be identified and showed good reliability.

Conclusion Further adaptations regarding measurement methods should be tested to gain better control of the kinetics

and kinematics of the EMG probe and accelerometers. To our knowledge this is the first study to test the reliability of PFM activity and time variables during dynamic functional whole body movements. More knowledge of PFM activity and time variables may help to provide a deeper insight into physical strain with high force impacts and important functional reflexive contraction patterns of PFM to maintain or to restore continence.

Keywords Jogging · Pelvic floor · Reproducibility · Stress urinary incontinence

Introduction

Stress urinary incontinence (SUI) is a widespread condition among women of every age, meaning that young female adults are also affected [1]. Brown et al. [2] investigated the prevalence, frequency, and severity of urinary symptoms by questioning nulliparous women aged ≥ 18 years during pregnancy: 10.8 % of the women reported urinary incontinence according to the authors’ definition, which consisted of experiencing leakage of urine at least once per month during the period of 12 months before pregnancy with their first child.

Among the young women suffering from SUI, young female athletes are also affected, and the highest prevalence is found in those involved in “high-impact” sports [3]. Carls [4] found that 28 % of the 14- to 21-year-old college and high school female athletes responding to a survey assessing the prevalence of SUI and the need for preventative urinary incontinence education reported symptoms of SUI during sports activities. More than 16 % of the responding female athletes with incontinence reported a negative effect on their quality of life that had an impact on their social life or desire to continue participating in sports [4]. Simeone et al. [5] assessed

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the prevalence of lower urinary tract symptoms (LUTS) and incontinence in female athletes. They found a prevalence of urinary incontinence of about 30 % (9.1 % SUI, 7.7 % mixed incontinence, 13.1 % urge incontinence) and also a more frequent association of high-impact sports with incontinence as opposed to low-impact sports with LUTS [5]. Eliasson et al. [6] investigated trampolining, which exerts extreme forces on the pelvic floor muscles (PFM) with every impact on the trampoline, and found that 80 % of the 35 trampolinists with a mean age of 15 years participating in the National Swedish Trampoline team reported urinary leakage during trampolining. All 28 subjects reporting urine leakage experienced the incontinence during trampoline training, and two subjects also reported leakage during other sports activities. None reported leakage during coughing, sneezing or laughing. However, not only female athletes suffer from SUI; Salvatore et al. [7] found that SUI affects a significant proportion of young women practicing non-competitive sports activities, and that it can cause sufferers to give up the sport or limit its practice.

Women who are physically active raise their intra-abdominal pressure more frequently than sedentary women [8], and according to Bø et al. [3] it is therefore debatable whether female athletes have strong PFM as a result of their regular training, thus preventing SUI, or whether excessive loading of the PFM may stretch and weaken them and cause SUI.

To guarantee continence, the PFM must be able to contract strongly [9], rapidly, and reflexively [10, 11]. The ability of PFM to generate rapid and strong contractions results in the generation of an adequate squeeze pressure in the proximal urethra, which maintains a pressure higher than that in the bladder, thus preventing leakage [12]. Rapid PFM contractions are crucial for maintaining continence preceding an abrupt rise in the intra-abdominal pressure associated with coughing and sneezing [11, 12]. Several studies showed that the PFM function regarding rate of force development was impaired in incontinent women compared with continent women [10, 11].

According to Hodges et al. [13], it is well known that muscles surrounding the abdominal cavity, such as the diaphragm and abdominal muscles, are active in association with tasks that challenge spinal stability. In their investigation of whether PFM contribute to the pre-programmed postural activity of the trunk muscles prior to predictable challenges to spinal stability, they found a mean onset of the PFM electromyography (EMG) activity of 28.4 ms prior to the onset of EMG activity of the deltoid during single fast arm movements [13]. Sjødahl et al. [14] found even earlier onsets of PFM activity relative to arm lifts and also showed a feed-forward response of the PFM to the start of a leg lift. To date there have been no such studies investigating PFM activity preceding movements of the whole body, as occurs, for example, during sports activities.

The highest prevalence of SUI is found in sports involving high-impact activities [3]. Running is one of the most popular physical activities worldwide. Zadpoor and Nikooyan [15], in their systematic review investigating the relationship between lower-extremity stress fractures and the ground reaction force, found peak ground reaction forces between 2.40 and 3.87 times the body weight. With increasing running speed, the ground contact gets shorter and the ground reaction force higher [16].

To date there have been no devices measuring the strength of the PFM during dynamic functional whole body movements. Consequently, EMG measurements are the only available appropriate method and therefore this investigation focuses on PFM activity during running by means of surface EMG. EMG in general is an often used and well-established method of measuring and investigating the voluntary or involuntary activity of skeletal muscles [17]. Electromyography during dynamic functional whole body movements allows the investigation of inter- and intramuscular coordination. Through determination of time variables of muscle activity, critical elements of muscle activity can be examined, e.g., precipitate or lagged activity [13, 14]. To the authors' knowledge there are no such studies investigating the time parameters of the PFM during dynamic functional whole body movements.

A better understanding of the PFM activity pattern during running would be relevant to the development of more specific diagnostic methods and therapy. Parameters such as PFM activity peak, time to activity peak ("EMG–time curve"), pre-contraction and time of pre-contraction (anticipation, feed forward), all in relation to the time of the heel-strike, could be crucial for gaining a better insight.

The aim of this study was to investigate PFM activity during running. The specific goals were to describe and test the reliability of the PFM EMG activity and time variables during running.

Materials and methods

Study design

The current investigation was designed as a prospective cross-sectional study to characterize PFM activity during running by means of vaginal EMG. It focuses on the intra-session test–retest reliability of time and activity variables derived from accelerometry and EMG. The study was conducted in accordance with the Declaration of Helsinki, and all subjects gave written informed consent. Following an unwritten agreement with the ethics committee, approval was not required as the investigation concerned a physiotherapy-relevant reliability low-risk study involving healthy volunteer students only and the investigation was a mandatory part of the physiotherapy study program.

Subjects

Ten female subjects from the physiotherapy bachelor program at Bern University of Applied Sciences, Health Section, were included on condition that they were aged between 20 and 35 years, were nulliparae, were anamnesticly healthy, had a BMI between 20 and 30 kg/m² and were physically able to cope with the requirements of treadmill running. Subjects who were pregnant, who were having their period or who had had surgery in the urogenital region as well as those with acute vaginal infection, incontinence, pelvic floor complaints, any pain during jogging, acute back or joint complaints or nickel or latex allergy were excluded. All subjects were familiar with the correct PFM contraction through their urogynecological lessons at the physiotherapy school. For the demographics of the subjects see Table 1.

Instrumentation

The running treadmill system used was a Quasar med (h-p-cosmos, Traunstein, Germany). All subjects had to perform their running at a velocity of 8 km/h and 1° inclination. A vaginal surface EMG probe (Periform®; Neen UK, Oldham, Lancashire, UK) was used to measure pelvic floor muscle activity. The differential adhesive surface electrode (Nicolet; Viasys Healthcare, Warwick, UK) was fixed on the right iliac crest. EMG was measured with a Tele Myo 2400 G2 device (Noraxon USA; Scottsdale, AZ, USA).

Three accelerometers (Model 317A 3-D accelerometers; Noraxon USA Inc) were attached with double- and single-sided adhesive tape (3 M and 3 M Durapor, Rueschlikon, Switzerland) on the lateral malleolus of the right leg to identify the time point of heel strike (T0), on the sacrum and on the external part of the vaginal probe to identify the vertical impacts at certain time points of the running cycle (see Table 2). The vaginal accelerometer was additionally coated with a latex finger stall (Sanor Lamprecht, Regensdorf, Switzerland).

Procedures

After emptying their bladder, receiving the test information and giving their written informed consent, the subjects were

equipped with the accelerometers and EMG reference electrode. The subjects were instructed in the vaginal insertion of the EMG probe using ultrasound lubrication and then performed the insertion themselves. The subjects wore a loose jogging suit and running shoes.

Electromyography was measured twice for 30 s while the subjects were at rest and totally relaxed, and twice for 5 s during maximum voluntary contraction in both a standing and supine position. Thereafter, the subjects performed a warm-up of walking (5 km/h) for 30 s, then running for 2 min 30 s in a steady state (8 km/h) to get familiar with the treadmill. After a short break of 1 min the treadmill was started again and accelerated up to 8 km/h. When the treadmill and the subject reached the final velocity of 8 km/h and were in a steady state the data acquisition was started: EMG and accelerometry were measured continuously for 60 s and the first 10 steps of these 60 s were analyzed. The subjects were instructed to run and to breathe as normally as possible and not to activate their pelvic floor muscles voluntarily and not to talk for 60 s.

Data reduction

Electromyography and accelerometer data were amplified with a gain of 500 and 1,000 and sampled at a rate of 2 kHz 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).

For the sampling frequency of 2 kHz, the sampling interval (dt) was 0.5 ms. The EMG signals were first low-pass filtered with a cut-off frequency of 1 kHz to avoid aliasing and high-pass filtered with 10 Hz (2nd order zero lag Butterworth filter, 24 dB/octave filter steepness). Second, EMG data were calculated as amplitude (µV) or as moving average (time window 200 ms) root-mean-square values and normalized to maximum voluntary contraction (MVC; (%)) EMG).

One hundred percent of EMG equals the average of the maximum values during the two times 5 s of the MVC. This MVC testing while standing was chosen instead of the usual MVC testing while in a supine position [18] because it seems more functional in comparison with running.

Accelerometer signals were low-pass (2nd order zero lag Butterworth) filtered with a cut-off frequency of 30 Hz.

The different time and activity variables are shown in Table 2. All variables were analyzed using the software package “Analoge und digitale Signalverarbeitung” (ADS) version 1.12 (uk-labs, Kempen, Germany).

Electromyography data at rest and at MVC were averaged over the test and retest, as were EMG data during running consisting of ten steps of the right leg for each subject.

Table 1 Subject demographics

Parameter	Unit	<i>n</i> or mean ± standard deviation
Included subjects	<i>n</i>	10
Age	Years	24.10±2.77
Weight	Kilograms	57.70±5.56
Height	Meters	1.67±0.07
Body mass index	kg/m ²	20.75±1.66

Table 2 Labels, units, and description of time and activity variables derived from accelerometry and electromyography (EMG)

Variable (unit)	Description	To identify
T0	Heel strike	The initial time point of the impact and beginning loading phase and strain of the PFM
Tsac (ms)	Time difference between heel strike (T0) and first acceleration of the sacrum accelerator	To identify the time difference (delay) between T0 and Tsac as a more bony part compared with the vaginal accelerator, which is attached to rather less rigid muscle structures
Tvagmin (ms)	Time difference between heel strike (T0) and minimal acceleration of the vaginal accelerator	Periodic characteristics of the acceleration of the vaginal sensor/PFM movements during cyclic running (from minimum to maximum)
Tvagmax (ms)	Time difference between heel strike (T0) and maximal acceleration of the vaginal accelerator	Periodic characteristics of the acceleration of the vaginal sensor/PFM movements during cyclic running (from maximum to minimum)
Temgmin (ms)	Time difference between heel strike (T0) and minimal EMG activity	Periodic characteristics and an “ON/OFF criterion” of the PFM activity in comparison to rest activity during supine position or standing
Temgmax (ms)	Time difference between heel strike (T0) and maximal EMG activity	Periodic characteristics of PFM activity
EMGmin ((%) EMG)	Minimal EMG activity	Periodic characteristics and an “ON/OFF criterion” of the PFM activity in comparison to rest activity during supine position or standing
EMGmax ((%) EMG)	Maximal EMG activity	Periodic characteristics of PFM activity
EMGmin-max ((%) EMG)	Mean EMG activity between EMGmin and EMGmax	The range of PFM activity
EMGT-50 ((%) EMG)	EMG activity at heel strike (T0) minus 50 ms	The pre-activity of PFM as a regulatory component of anticipation
EMGT0 ((%) EMG)	EMG activity at heel strike (T0)	The EMG amplitude at the initial time point of the impact and beginning the loading phase and strain of the PFM
EMGT-50-T0 ((%) EMG)	Mean EMG activity between minus 50 ms and T0	The mean pre-activity
EMGmin-max (reg) ((%/s) EMG)	Linear regression of EMG activity between EMGmin and EMGmax	The gradient of range of amplitude
EMGT-50-T0 (reg) ((%/s) EMG)	Linear regression of EMG activity between -50 ms and T0	The gradient of pre-activity

Statistical analysis

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

The reliability test followed the three steps suggestions of Weir [19]: In order to identify possible systematic errors between the repeated measures, the Wilcoxon rank order test was applied at rest and the MVC tests, and the Friedman test to compare the dependent time and EMG variables over the ten steps. Intra-session test–retest reliability (consistency) was calculated with the intraclass correlation coefficient (ICC 3.1; i.e., relative reliability) and the standard error of measurement (SEM; i.e., absolute reliability). Additionally, the minimal detectable differences (MDDs) using a 95 % confidence interval and the mean absolute differences among all repeated measures were calculated.

Portney and Watkins [20] suggest as a general guideline that ICC values above 0.75 are indicative of good reliability and those below 0.75 of poor to moderate reliability.

The level for significances was set at $P < 0.05$. All statistics were calculated using SPSS 17 for Windows (SPSS; Chicago, IL, USA) as well as Microsoft Excel 2007 (Microsoft, Redmond, WA, USA).

Results

Figure 1 shows samples of the amplitude and time variability of EMG–time curves within the ten steps of one subject.

Thirteen variables were selected based on ten steps of each subject. Descriptive statistics and reliability calculations are presented in Tables 3 and 4. All baseline data of EMG during rest or MVC show good reliability. The analysis of systematic errors within repeated measures by Wilcoxon or Friedman tests revealed only nonsignificant values for all variables. ICCs for the time variables (ms) ranged from 0.113 to 0.731. The highest reliability indexes were found for all EMG-amplitude-derived variables

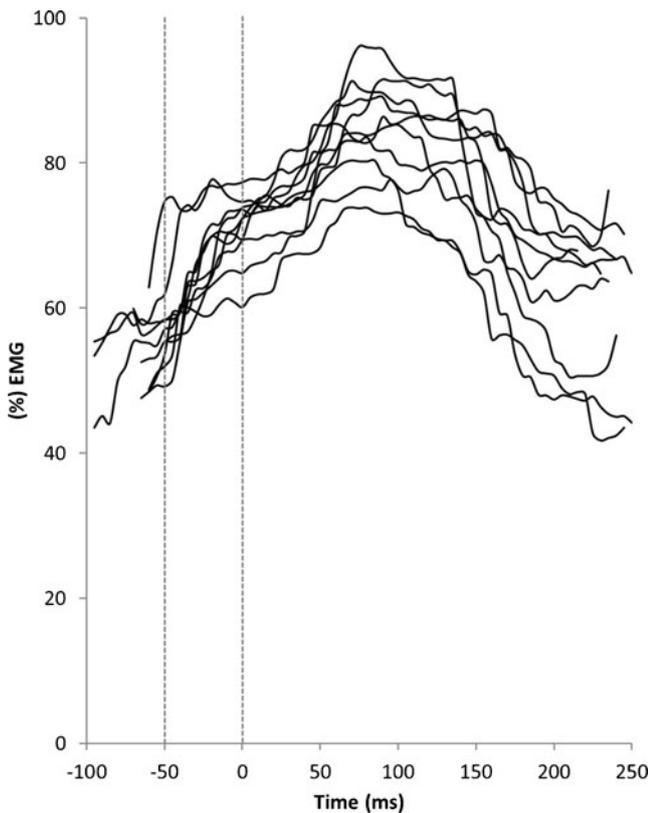


Fig. 1 Exemplary EMG–time curves derived from the ten steps of subject 1. The curves start and end at the minimum EMG (EMGmin) activity of each step cycle. The *dotted line* at -50 ms (EMGT-50) represents the start of the pre-activity phase which ends at T0 (*dotted line* at 0 ms = heel strike). Subsequently, EMG–time curves increase to their activity maximum (EMGmax)

(%) EMG), which were superior to 0.750 and ranged between 0.906 and 0.942. The combined time–amplitude variables ((%/s) EMG) did not reach excellent reliability, with values of 0.542 and 0.704.

Discussion

The purpose of this study was to describe and reliability test the variables that characterize PFM activity and time behavior during running. We selected a total of 13 variables (see Table 2), which were tested with regard to their intra-session retest reliability based on ten steps of each subject. Generally, EMG variables showed good time variables and poor reliability.

EMG variables

Among the technologies used to study PFM neural control, EMG of the PFM has gained the most attention and clinical relevance [21]. Grape et al. [18] tested the retest reliability of isolated PFM contraction exercise by means of EMG in healthy nulliparae women and showed good reliability

(ICC=0.82–0.96) of average activity (during a 10-s long squeeze), peak (maximum value), work (the area under the curve), and baseline (resting value); the ICC for speed (peak divided by time taken to reach the peak) was between 0.65 and 0.85. In contrast to our investigation, Gollhofer et al. [22] investigated the EMG reliability in leg muscles during running at constant velocity. They found high reliability coefficients for day-to-day and week-to-week comparisons and also, with regard to qualitative comparisons of EMG, high reproducibility of the shape of the patterns. However, to date there have been no reliability studies concerning vaginal EMG during dynamic functional whole body movements of the subjects.

In our study 6 of the 8 identified EMG variables of PFM activity during running showed good ICC values. Moderate to almost good ICCs of EMG measurements were found for regressions.

The minimal EMG (EMGmin) activity during running shows good reliability. Mean minimal EMG activity during running shows an activity about 24 % higher than mean PFM EMG activity during rest in a standing position. These findings show a constantly increased PFM tone during running, and an aerobic strain of PFM during running could be hypothesized. The increased tonicity of PFM activity was also found in the study of Hodges et al. [13] during repetitive arm movements.

The maximal EMG (EMGmax) activity shows good reliability. The mean activity level was at 124.3 (%) EMG. Hodges et al. [13] found, during repetitive arm movements, bursts of anal and vaginal EMG activity in association with the frequency of arm movement, but they did not provide information about the activity level. These findings suggest a higher PFM activity compared with MVC owing to reflexive and reactive force generation during running.

The mean EMG activity between EMGmin and EMGmax variables shows good reliability and a mean level of 99.2 (%) EMG. These values are about 68 (%) EMG higher than during standing at rest. Schulte-Frei [23] examined PFM activity while subjects stood on stable and unstable ground and during one-leg stands. As her MVC normalization referred to an MVC taken individually during co-contraction while performing a set of other exercises for muscle testing, these values cannot be compared with those found in our study. Nonetheless, as she found increasing PFM mean activity from standing on stable ground compared with standing on unstable ground and an even higher activity during the one-leg stands, these findings are of interest as, during running a temporary one-leg stand takes place as well.

The variable linear regression of EMG activity between EMGmin and EMGmax shows almost good reliability.

The EMG activity at heel strike (T0) minus 50 ms shows good reliability and an activity level of 72.1 (%) EMG, which is about 40 (%) EMG higher than during rest. Hodges et al. [13] also found a mean PFM pre-activity of

Table 3 Descriptive statistics (mean \pm SD), reliability indexes (intraclass correlation coefficient, standard error of measurement), minimal detectable difference for variables, and test for systematic error (Wilcoxon, difference) for activity variables

Variable	Units	Mean	SD	ICC	SEM	MDD	Wilcoxon P value	Difference Mean
Rest supine	(%) EMG	14.2	9.7	0.999 ^a	3.0	8.3	0.398	0.1
Rest standing	(%) EMG	31.3	4.0	0.993 ^a	4.0	11.1	0.173	0.7
MVC supine	μ V	64.7	27.4	0.867 ^a	8.7	24.1	0.445	1.3
MVC standing	μ V	57.6	20.1	0.914 ^a	6.3	17.6	0.508	1.1

Mean arithmetic mean, SD standard deviation, ICC intraclass correlation coefficient, SEM standard error of measurement, MDD minimum detectable difference, Wilcoxon Wilcoxon signed-rank test; Difference arithmetic mean of test–retest differences

^a Indicates ICCs > 0.750

28.4 ms prior to arm movements and Sjødhall et al. [14] a median PFM pre-activity of 206 ms prior to the start of a leg lift at a level of 20 (%) EMG. These findings confirm the assumption that the increased PFM activity 50 ms prior to the heel strike could indicate reliable PFM pre-activity during running.

The EMG activity at heel strike (T0) shows good reliability. As the PFM activity shows higher values than the activity at rest in a standing position, this variable confirms the assumption of a PFM pre-activity.

The mean EMG activity between minus 50 ms and T0 shows good reliability. Again, the values are higher compared with the PFM activity during rest in a standing position, which could indicate PFM pre-activity.

The linear regression of EMG activity between –50 ms and T0 before the heel strike shows moderate reliability. This value

((%/s) EMG) cannot be used to investigate PFM pre-activity, as it cannot be compared with the PFM MVC ((%) EMG).

Time variables

The time variables show large ranges and therefore low reliability. Four (time difference between heel strike (T0) and minimal acceleration of VagAcc; time difference between heel strike (T0) and maximal acceleration of VagAcc; time difference between heel strike (T0), and minimal EMG activity; time difference between heel strike (T0) and maximal EMG activity) out of five time variables showed low ICC values.

The only variable showing almost good reliability was the time difference between heel strike (T0) and first acceleration of SacAcc. A possible explanation could be a more precise force transmission on bony structures than on less rigid muscle structures.

Table 4 Descriptive statistics (mean \pm SD), reliability indexes (ICC, SEM), MDD for variables, and test for systematic error (Friedman, difference) for time and activity variables derived from accelerometry and electromyography

Variable	Units	Mean	SD	ICC	SEM	MDD	Friedman P value	Difference Mean
Tsac	ms	72.3	14.8	0.731	4.7	13.0	0.751	0.4
Tvagmin	ms	–10.0	42.3	0.169	13.4	37.0	0.628	1.4
Tvagmax	ms	75.5	13.0	0.113	6.6	18.4	0.694	3.2
Temgmin	ms	–3.7	57.2	0.452	18.1	50.1	0.604	1.9
Temgmax	ms	214.2	51.8	0.390	16.4	45.4	0.517	–2.3
EMGmin	(%) EMG	55.0	24.9	0.906 ^a	7.9	21.8	0.311	–1.6
EMGmax	(%) EMG	124.3	99.2	0.937 ^a	31.4	87.0	0.848	5.6
EMGmin-max	(%) EMG	99.2	81.6	0.942 ^a	25.8	71.5	0.796	2.9
EMGT-50	(%) EMG	72.1	40.1	0.935 ^a	12.7	35.2	0.064	–1.5
EMGT0	(%) EMG	64.0	27.0	0.917 ^a	8.5	23.6	0.164	–1.6
EMGT-50-T0	(%) EMG	66.6	31.3	0.942 ^a	9.9	27.4	0.226	–1.3
EMGmin-max (reg)	(%/s) EMG	155.9	131.7	0.704	41.6	115.4	0.494	16.7
EMGT-50-T0 (reg)	(%/s) EMG	–34.6	98.4	0.542	31.1	86.3	0.709	–13.1

Mean arithmetic mean, SD standard deviation, ICC intraclass correlation coefficient, SEM standard error of measurement, MDD minimum detectable difference, Friedman Friedman test nonparametric repeated measures comparisons; difference arithmetic mean of test–retest differences

^a Indicates ICCs > 0.750

To our knowledge, there has been to date no literature published on the time variables of PFM contraction characteristics, and therefore the results of this study can only be discussed in the context of other findings. Leitner et al. [24] were able to show good reliability regarding time variables in stair ascent in the elderly, but clearly lower reliability in stair descent, which seems to be more comparable to impacts during running at slow velocity. Masani et al. [25] investigated the variability of the ground reaction force pattern during walking in relation to different constant speeds. He found an increasing trend in variability with speeds from 3 to 8 km/h; however, the minimum variability speeds of 5.5–5.8 km/h were within the limits of usual walking speeds. This result indicates that the neuromuscular locomotor system was most stable at usual walking speed.

Methodical problems of vaginal probe measurements

Because vaginal probes cannot be fixed directly onto the muscles, as they use self-adhesive electrodes, minimal shifts of vaginal probes might go unnoticed. In the present study the attachment with tape of the accelerometer to the vaginal probe seemed to be sufficient. However, greater movement of the vaginal probe (EMG and accelerometer) in contrast to the sacral time variables (Tsac) cannot be totally excluded and is supported by the lower ICCs of the vaginal time variables (Tvagmin and Tvagmax). In a future study this issue could be addressed by using a device such as a STIMPON® electrode (Innocept Biobedded Medizintechnik, Gladbeck, Germany). This electrode adapts its shape to individual vaginal cavities and therefore movement could be avoided.

Cross-talk

Electromyography is a reliable method of assessing PFM activity in healthy women [18]. However, Bo and Sherburn [26] conclude in their overview assessing PFM function and strength that the EMG signals must be interpreted with caution because the risk of cross talk from other muscles is high and because of the variability in electrode placement within the vagina. To date, cross-talk during functional dynamic whole body movements for PFM activity has not yet been examined and is therefore difficult to estimate.

Treadmill

Some subjects were not familiar with treadmill running, which possibly resulted in a nonphysiological gait compared with running on natural ground at usual individual speeds. However, running on natural ground at individual speeds would be problematic with regard to speed standardization. To address the problem of a lack of familiarity with treadmill running, all the subjects performed a warm-up of

5 km/h for 30 s and then a running-in of 8 km/h for 2.5 min, both on the treadmill, before the measurements took place.

Shoes

According to Maurer et al. [27], the characteristics of human locomotion may change for different boundary conditions such as footwear. In our study, the subjects wore their own running shoes. Wright et al. [28] found no difference between soft and hard shoe impact forces. However, peak rates of loading were greater for the hard shoe than for the soft shoe. Therefore, an influence of footwear on gait and force transmission cannot be excluded. Future studies should require standardized running shoes or running barefoot.

Conclusion

To the best of our knowledge, the current study is the first to test PFM activity and time variables during dynamic functional whole body movements with regard to their reliability. It provides a first insight into the PFM activity of healthy women during running. EMG variables of good reliability could be shown, while poor evidence was gained for the reliability of time variables. In particular, further studies should consider adaptations regarding the vaginal probe and footwear. The test application and reliability will likely have to be modified and retested in different populations and given the anatomical changes seen associated with SUI and pelvic organ prolapse. More knowledge of PFM EMG variables and time variables may help to provide a deeper insight into physical strain with high force impacts and functionally important reflexive contraction patterns of PFM to maintain or to restore continence. A further consequence is support for the development of more sophisticated diagnostic instruments and therapeutic approaches.

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Conflicts of interest None

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