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Published in:
Journal of Electromyography & Kinesiology

DOI (link to publication from Publisher):
10.1016/j.jelekin.2011.03.001

Publication date:
2011

Document Version
Early version, also known as pre-print

Journal of Electromyography and Kinesiology 21 (2011) 638–644

Contents lists available at ScienceDirect

Journal of Electromyography and Kinesiology

journal homepage: www.elsevier.com/locate/jelekin

Inverse relationship between the complexity of midfoot kinematics and muscle activation in patients with medial tibial stress syndrome

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ARTICLE INFO

Article history:
Received 4 November 2010
Received in revised form 4 March 2011
Accepted 4 March 2011

Keywords:
Gait
Variability
Entropy
Overuse injury
Complexity

ABSTRACT

Medial tibial stress syndrome is a common overuse injury characterized by pain located on the medial side of the lower leg during weight bearing activities such as gait. The purpose of this study was to apply linear and nonlinear methods to compare the structure of variability of midfoot kinematics and surface electromyographic (SEMG) signals between patients with medial tibial stress syndrome and healthy controls during gait.

Fourteen patients diagnosed with medial tibial stress syndrome and 11 healthy controls were included from an orthopaedic clinic. SEMG from tibialis anterior and the soleus muscles as well as midfoot kinematics were recorded during 20 consecutive gait cycles. Permuted sample entropy and permutation entropy were used as a measure of complexity from SEMG signals and kinematics

SEMG signals in patients with medial tibial stress syndrome were characterized by higher structural complexity compared with healthy controls ($p < 0.001$) while it was the opposite for the midfoot kinematics ($p = 0.01$).

Assessing the complexity of midfoot kinematics and SEMG activation pattern enabled a precise characterization of gait in patients with medial tibial stress syndrome. The reported inverse relationship in foot kinematics and SEMG complexity most likely point towards separated control processes governing gait variability.

Introduction

Medial tibial stress syndrome (MTSS) is a common overuse injury. Studies have reported a prevalence in select physically active populations ranging from 4% to 35% (Almeida et al., 1999; Burne et al., 2004; Plisky et al., 2007; Yates and White, 2004). MTSS are characterized by continuous or intermittent pain on the medial side of the lower leg. The pain can be dull or intense and is exacerbated during repetitive weight bearing activities (Kortebein et al., 2000).

As MTSS is primarily observed in subjects participating in weight bearing activity, kinematics of the lower extremity has been investigated in several studies (Bandholm et al., 2008; Bennett et al., 2001; Willems et al., 2006). Much attention has been paid to the kinematic characteristics of the foot and ankle. More specifically, descriptors of foot movement such as average dynamic navicular drop (dNH) have been associated with MTSS (Bennett et al., 2001; Bandholm et al., 2008).

Adequate function and activity of the leg muscles are considered as the needed pre-requisite to initiate the necessary biomechanical force for propulsion and decrease the impact to the tibia during walking or running (Fyhrie et al., 1998; Richie et al., 1993; Scott and Winter, 1990). Clement (1974) suggested that weakness of the calf muscles may actually predispose individuals to MTSS by leading to excessive force being transferred to the tibia. Surprisingly, surface electromyography (SEMG) from lower leg muscles has not yet been thoroughly investigated in patients with MTSS. Muscle activation profiles are generally quantified by computing temporal estimators such as root mean square or averaged rectified value. However, linear approaches may not be able to detect subtle changes in the neuromuscular system (Vaillancourt and Newell, 2002). Patients with MTSS are physically active, and salient changes in the neuromuscular system may take place. Thus, the assessment of the variability of kinematics and muscle activation during gait can be a way of assessing important aspects of motor control (DiBerardino et al., 2010; Latash et al., 2002). The magnitude of variability is measured by linear methods applied to time series, but nonlinear analysis is required to investigate the structure of variability. The structure of variability provides information on the deterministic and stochastic organization of the signal investigated. Nonlinear variability may be quantified by computing entropy measures. Among them, entropy

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doi:10.1016/j.jelekin.2011.03.001

measures are considered particularly suitable for the analysis of biological signals (Pincus et al., 1991; Richman and Moorman, 2000). Entropy quantifies the predictability of the signal investigated with higher value indicating a more complex structure and lower predictability of the time series.

Changes in the complexity of motor patterns may be related to changes in motor strategies and may thus reveal the effects of adaptations and pathologies (Bartlett et al., 2007). A number of authors (Goldberger 1997; Pool 1989; Vaillancourt and Newell, 2002) have proposed that a low complexity would characterize patient population while Stergiou et al. (2006) have challenged this concept and suggests that increases beyond optimal variability can also make the system noisy and unstable and may be associated with pathology (Stergiou et al., 2006).

Nonlinear analysis has not extensively been performed on SEMG signals (Samani et al., 2010a; Sung et al., 2005), but a few studies indicate that nonlinear methods may be more sensitive towards changes in neuromuscular system compared to linear methods (Fattorini et al., 2005; Felici et al., 2001). Meigal et al. (2009) computed sample entropy from SEMG signals from biceps brachii during different loads in subjects with Parkinson disease. At low contraction levels, patients with Parkinson's disease were characterized by slightly higher SaEn compared with young control subjects as opposed to the general hypothesis associating low complexity with pathology. Interestingly, Morrison et al. (2007) reported an inverse relationship between complexity of muscle pattern and centre of pressure dynamics during voluntary and random sway motions. These conflicting results regarding complexity and healthiness of biological signals is further supported by the possibility that distinct mechanisms are governing the neuromuscular and force variability (Sosnoff and Newell, 2006). Thus, there is a need for further investigation aiming at delineating the relationships between the complexity of SEMG and kinematics time series in patient populations, e.g., patients with MTSS.

The purpose of this study was to compare the structure of variability of dNH as well as the SEMG-signal from tibialis anterior and soleus between patients with MTSS and healthy controls. We hypothesized that patients with MTSS would be characterized by lower complexity of dNH and higher complexity of calf muscles activation compared with healthy controls.

Methods

Subjects

Fourteen patients with MTSS were included from an orthopaedic outpatient clinic. All patients were diagnosed with MTSS within a period of four weeks. The patients were diagnosed by an experienced physiotherapist together with an experienced orthopaedic surgeon. MTSS was defined as continuous or intermittent pain in the medial tibial region, exacerbated with repetitive weight-bearing activity, and localized soreness along the distal two thirds of the posterior-medial tibia crest (Kortebein et al., 2000). All patients had experienced symptoms for at least three months. The control group consisted of 11 healthy participants, Table 1. Subjects in the control group were recruited through advertising and consisted of university staff and students. All participants gave informed consent, and the study was approved by the Ethics Committee of North Denmark Region (N-20080052).

Measurements and instrumentation

In subjects with bilateral MTSS, all measurements were done on the most painful leg. The proportion of right and left legs was balanced between the two groups. The six-component Foot Posture Index (FPI-6) (Redmond et al., 2006) was used to classify the standing foot posture. The FPI-6 score is based on clinical observations consisting of five visual assessments of foot posture and a palpatory evaluation of the position of talus. The assessment on the subjects was carried out by the same practitioner and categorized using the terminology used by Redmond et al. (2006). Minus twelve to minus one was defined as a supinated foot posture, 0 to +5 was defined as normal foot posture and +6 to +12 was defined as a pronated foot posture. The preferred individual overground walking speed was used as the set speed for walking on the treadmill (Jordan et al., 2007). To determine this parameter, participants walked along a five meter runway with light sensitive start and stop timers in both ends. Gait speed was averaged from three trials. A visual analogue scale (VAS) with a length of 100 mm was used to quantify pain intensity during activity in both the control and case group.

Table 1
Demographic data from the group of patients with MTSS and control group. Values are mean ± 1SD. VAS rest and VAS activities are pain scores on a 0–10 scale presented as median and interquartile range.

Group characteristics	MTSS (n = 14)	Control (n = 11)
Age [years]	27.8 (8.8)	27.3 (6.2)
Height [cm]	172,7 (10.4)	174.5 (10.7)
Weight [kg]	77.8 (16,2)	76.1 (14.1)
BMI [kg/m^2]	25.8 (5.1)	24.9 (3.1)
VAS rest [cm]	0 (0–1)	0
VAS activity [cm]	6.5 (5–7)	0
Foot length [mm]	251 (18.6)	248 (17.0)

Data acquisition and processing

A custom-designed three-dimensional multi video sequence analysis system was used to automatically identify markers on the medial and posterior aspect of the foot during walking on a treadmill. The system consisted of two digital video cameras (Basler Scout, frame rate 86 Hz) both equipped with a 12 mm lens. One camera was placed perpendicular to the orientation of the treadmill with a distance of 2975 mm from the centre of the treadmill. A second camera was placed 3440 mm behind the treadmill. Two powerful light sources were placed close to the video cameras in order to get good reflection from the reflective markers. Data were transferred to a computer via firewire interface (StreamPix, Norpix, Montreal, Quebec, Canada). Three flat retro-reflective markers (diameter 12 mm) were placed on the medial side of the foot (marker 1, 2 and 3). For marker attachment, participants were asked to stand in a relaxed position with straight knees and their weight distributed equally to both feet. The feet were placed with their medial borders parallel and 15 cm apart. Using a laser alignment device (Stanley Works, New Britain, CT, USA) the markers at the first metatarsal (marker 1) and calcaneus (marker 3) were positioned 1.9 cm above the floor. Marker 2 was placed on the navicular tuberosity (Rathleff et al., 2010a).

Bipolar surface electrodes (Ambu A/S, Neuroline, Ballerup, Denmark) were placed on the m. tibialis anterior and m. soleus using the recommendations of SENIAM project (Hermens et al., 2000). The SEMG signals were amplified 2000 times and band-pass filtered (5–1000 Hz). EMG signals were sampled at 2 kHz, by a 12 bit A/D converter (Biovision, Werheim, Germany), and stored on a disk. Prior to analysis, the SEMG signals were digitally band-pass filtered (10–400 Hz, 4th order Butterworth). Furthermore, a notch filter (2nd order Butterworth band stop with rejection width 1 Hz centred at first four harmonics of the power line frequency (50 Hz)) was used when necessary to remove line interference. Foot-switches were taped to the foot and used to define gait phas-

ing as onset/offset of heel-strike and toe-off. Two foot-switches were taped to the sole of the foot, located under the heel and the great toe. The foot-switches were sampled synchronously with SEMG signal at 2000 Hz and stored on disk.

After preparing the setup and placing reflective markers and surface electrodes, the subjects walked for 6 min at the preferred overground walking speed (Matsas et al., 2000). Subsequently, the subjects walked on the treadmill for 20 s approximately equivalent to 20 consecutive steps. After additional 60 s of continuous walking, 20 s of SEMG was recorded.

Navicular height was calculated as the perpendicular distance between marker 2 and the line between markers 1 and 3. Subsequently the distance between the floor and the line between marker 1 and 3 were added.

In a pilot study this system showed good intratester reliability for measuring dND (ICC (2,1) within day: 0.89 and ICC between day: 0.94 (Rathleff et al., 2010a). Using a moving wooden leg with a fixed marker setup, precision and accuracy of the system was both determined to be 0.3 mm for navicular height (Jensen and Juhl, 2009).

Data analysis

Patients with MTSS only reported pain during weight bearing activity. Thus, we only analysed the stance phases of the gait cycle.

SEMG data were time normalized and linearly interpolated to 1000 points per stance phase of each stride, triggered by the heel-strike to toe-off. SEMG samples of the stance phase from each stride was divided into five identical time segments representing: (0–25–50–75–100%) of stance phase of each stride. RMS of SEMG over each time segment was computed and normalized to the maximum of SEMG RMS during stance. Then, the normalized values were averaged over all recorded strides for each time segment (Fig. 1).

The stance phase of all gait cycles were divided into five intervals representing (0–25–50–75–100%) of stance phase on all strides. Likewise, the dND data divided into two intervals representing (0–100%) stance phase on all strides.

Permuted sample entropy (Samani et al., 2010b) (PeSaEn) and permutation entropy (PE) (Bandt and Pompe, 2002) were calculated from the dND and SEMG data for each one of the aforementioned intervals during stance phase. Entropy quantifies the complexity of a dataset by assessing the probability that equal sequences of length *m*, will remain similar after a time increment. The output is a unitless, non-negative number where lower values indicate more regular signal and higher values more complex signal (Harbourne et al., 2009; Madeleine and Madsen, 2009; Richman and Moorman, 2000). In this study, we computed PE and PeSaEnas these two have been found to be a valid estimate of time series complexity (Bandt and Pompe, 2002; Samani et al., 2010b). PeSaEn values were calculated from the dND and SEMG time series for each one of the aforementioned time intervals during stance phase. The embedding dimension, m, and the tolerance distance, *r*, of navicular height were chosen on the basis of previous human movement studies (Madeleine and Madsen, 2009; Stergiou 2004; Vaillancourt and Newell, 2000).

Statistical analysis

Two-way ANOVA was applied for PeSaEn and PE values introducing time and subject groups (healthy, MTSS) as factors in analysis of SEMG data. Similar, one-way ANOVA analysis was performed for normalized RMS separately for each time interval (0–25–50–75–100%) of stance phase. The size of dND was compared using Student's t-test without including time as a factor. Mean (SD) are reported and $p < 0.05$ was considered significant.

Sample size calculations were conducted a priori based on the size of navicular drop. Minimum relevant difference was set as 2.0 mm and the level of significance was set at $p < 0.05$. The variance was set at 1.9 and was based on results from a previous study (Nielsen et al., 2010). With a statistical power of 80%, calculations revealed that a sample of 14 subjects in each group was re-

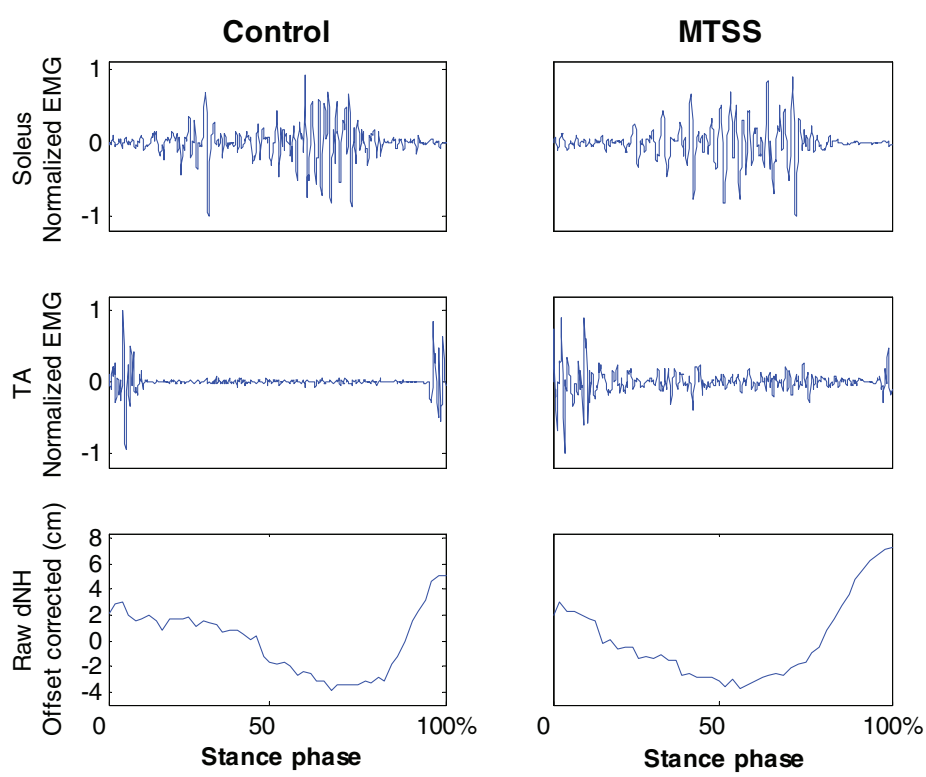

Fig. 1. EMG samples of stance phase of one stride for one representative subject normalized to the maximum of EMG in that stride in soleus and TA.

quired. However, three subjects in the control were excluded as the EMG recordings were not suitable for analysis.

Results

Ten of the 14 subjects with MTSS experienced bilateral symptom during weight bearing activity, six of these experienced the worst pain on the left side and eight on the right side. There was no difference in preferred overground walking speed among groups ($p = 0.56$). Subjects with MTSS had on average a FPI of 3.1 (1.6) and controls had a FPI of 3.3 (1.8), ($p = 0.77$). Similarly, no significant difference was found in the FPI score among groups ($p > 0.05$). During treadmill walking, the patients with MTSS had, on average, a 1.5 mm larger dND, ($p = 0.004$) compared with the control group. There was no significant difference in standard deviation of dND between groups ($p = 0.27$). Finally, dND showed a less complex pattern among the MTSS patient group compared with healthy subjects as depicted by PeSaEn values ($F_{1,46} = 8.0$, $p = 0.007$) (Fig. 2).

Subject groups played a significant role on normalized RMS for TA at the first time interval (0–25%) corresponding to the initial part of contact time ($F_{1,23} = 4.48$, $p = 0.04$). Normalized RMS was higher for healthy subjects (64 (8)%) compared with the MTSS group (54 (12)%). Subject groups also tended to play a significant role at the last time interval corresponding to 100% of contact time ($F_{1,23} = 3.6$, $p = 0.07$). In general, the level of muscular activation of the TA tended to be higher for healthy subjects (25 (10)%) compared with MTSS group (18 (9)%).The activity of Soleus and TA were more complex among the MTSS patient group as depicted by higher PE values for Soleus and TA ($F_{1,115} = 6.9$, $p = 0.01$ and $F_{1,115} = 5.5$, $p = 0.02$). PeSaEn showed no significant difference between groups for TA ($F_{1,115} = 1.6$, $p = 0.20$) (Fig. 3).

Discussion

This study confirmed that patients with MTSS with pain in the tibial region are characterized by lower complexity in midfoot kinematics, but also by higher complexity of the SEMG signal of the tibialis anterior and soleus when compared to a group of healthy controls.

Stance phase characteristics in MTSS patients

The patients with MTSS were characterized by a 1.5 mm larger navicular drop and a lower complexity of the movement of the navicular on average. In a previous study, we have reported that a larger navicular drop is associated with low complexity, and low movement of the midfoot is associated with high complexity (Rathleff et al., 2010b). Despite the relationship between the size of the navicular drop and complexity of the midfoot kinematics, not all subjects with a large navicular drop also had a low degree of complexity (Rathleff et al., 2010b). Most likely, this indicates that navicular drop and the structure of kinematics variability may represent different constructs as suggested recently (Moe-Nilssen et al., 2010).

Pool (1989) has proposed the "loss of complexity" hypothesis stating that healthy subjects have a certain amount of innate variability and that a loss of this variability will characterize pathology. Later investigations have confirmed this hypothesis (Sung et al., 2005; Vaillancourt and Newell, 2002). However, the causative link between low movement variability and injury is still unclear. Hamill et al. (1999) examined variability of knee joint coupling in patients with patellofemoral pain syndrome and compared them to healthy controls. They reported that healthy subjects display greater efficiency and variation in the movement around the knee joint when compared to patients with patellofemoral pain syndrome. Based on their results, they hypothesized that the low variability of the knee joint couplings in the symptomatic individuals indicates repeatable joint actions within a very narrow range. This narrow range of motion in combination with a low variability could result in a uniform stress on tissue surrounding the knee and thereby causing injury in the long run. Their results are strengthened by the work of Donoghue et al. (2008) who examined runners with a history of chronic Achilles tendon injuries and compared them with healthy controls with no history of injury. The group with a history of injury had a significantly lower variability of rear foot movement in the first half of the stance phase. These kinematic findings are in line with the results from the present study, all indicating that reduced complexity of kinematics is one of the intrinsic characteristic of MTSS patients. One of the muscles that might influence foot kinematics during heel strike and through the stance phase is the tibialis posterior. Fatigue and decreased force output of the tibialis posterior does not seem to affect movement rear foot or forefoot (Pohl et al., 2010), but interestingly it appears fatigue of the tibialis posterior has a significant effect on joint coupling between the foot and shank and the variability of joint coupling (Ferber and Pohl, 2011). One might speculate that tibialis posterior and the peroneus longus and brevis

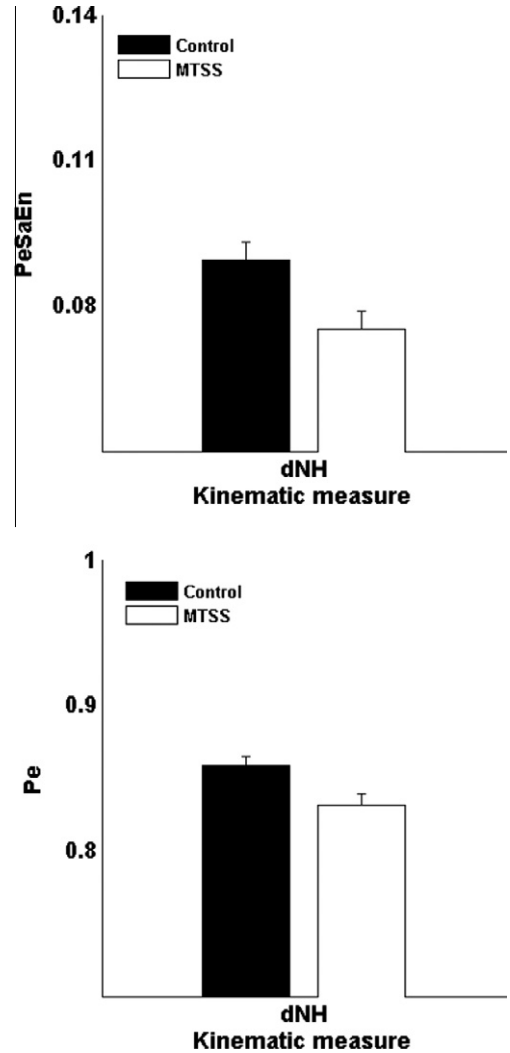

Fig. 2. Mean (SE) Permuted sample entropy (PeSaEn) and Permutation entropy (Pe) of dNH comparing the healthy control group versus MTSS patient group during gait.*$P < 0.05$.

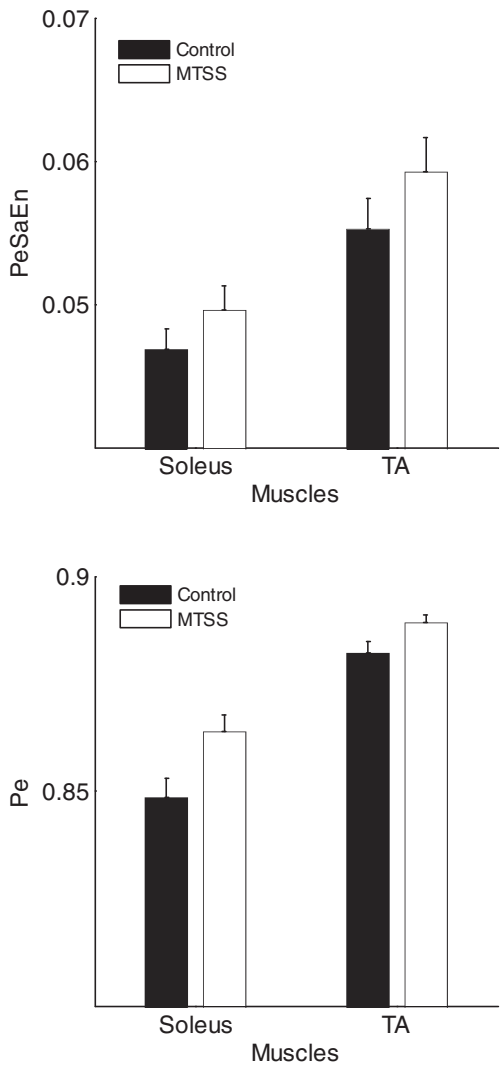

Fig. 3. Mean (SE) Permuted sample entropy (PeSaEn) and Permutation entropy (Pe) of electromyographic activity of Soleus and tibialis anterior muscles comparing the healthy control group versus MTSS patient group during gait. $^{*}P < 0.05$.

both might function as stabilizers at the medial and lateral part of the subtalar joint, and thereby stiffen the joint during heel strike and loading which might influence complexity from mid foot kinematics.

The overall activity for both the soleus and tibialis anterior tended to be slightly higher for healthy subjects and was significantly higher at the start of the stance phase for the soleus muscle. Clement (1974) suggested that decreased muscle strength in the lower leg could cause a larger impact through the tibia during walking or running. This idea is on a par with our results as patients with MTSS tended to display a lower activity in the soleus and tibialis anterior underlining most likely a protective mechanism towards pain. One could speculate further that a decrease in muscle activation during heel strike will contribute to increase the velocity of foot movement and thereby increase the forces transmitted through the tibia during heel-strike.

PeSaEn and PE were computed to assess differences in the structure of variability of the SEMG and kinematics time series. Entropy measures have been used extensively to examine the degree of complexity of, e.g., electroencephalograms (Natarajan et al., 2004), electrocardiograms and heart rate variability (Pincus et al., 1993) and centre-of-pressure measurements (Morrison et al., 2007). The use of nonlinear technique to examine the SEMG signal and kinematics provides additional insight into the complex interaction between kinematics and muscular activity. In this study, we found differences in the structure of the SEMG signal between MTSS patients and controls, i.e., higher SEMG complexity for patients. Contrary to us, Sung et al. (2005) found that patients with low back pain had a lower structure of variability of erector spinae during a fatiguing task compared to healthy subjects. Lamoth et al. (2004) investigated how experimentally induced pain affected SEMG from erector spinae and trunk kinematics and reported an increase in SEMG variability in acute pain condition in line with our results. Acute experimental muscle pain (intra-muscular injection of hypertonic saline) induced in the dominant tibialis anterior induced a reorganization of muscle synergies (Madeleine et al., 1999) mostly inline with the present findings. However, MTSS is often a chronic condition with a long duration of symptoms difficult to replicate by means of experimental pain models. In chronic low back patients, gait is characterized by both less and more variable kinematic coordination in the transverse and frontal plane, respectively, together with poor lumbar erector spinae muscles coordination (Lamoth et al., 2006). Most likely, the differences in patient population and in the type of motor task explain these discrepancies (Madeleine et al., 2011).

In this study, the healthy subjects exhibited increased complexity in their kinematics associated with a decrease in SEMG while the opposite pattern was found among patients with MTSS. As greater SEMG signal irregularity most likely reflects less predictable muscle firing patterns, one would have expected that the most complex SEMG signals would be paired with the healthy subjects. However, a few studies have actually reported inverse relationship among center of pressure measurements and kinematics (Morrison et al., 2007). These findings can be interpreted using the conceptual framework of complexity tradeoffs between the macroscopic and microscopic levels of a system (Bar-Yam, 2004). Such tradeoffs are reflective of changes in the organization of the degrees of freedom within the system through changes in the level of independence between their behaviour patterns. Thus, the number of attractors used at a muscle level is likely to differ from the ones used in kinematics of gait. Such a relationship suggests that the level of complexity in the neuromuscular systems may be limited and is inversely proportional to the complexity at the kinematics level.

Thus, studies that only investigate SEMG or kinematics may not be able to investigate complexity tradeoffs between different subsystems. The results of the present study argue for independent mechanisms governing variability in muscular activity and the kinematics of the foot during gait.

Limitations

Because of the case-control design of this study, no cause-and-effect relationship can be inferred from our results. With the current sample size there were no significant group difference in static foot posture between subjects with MTSS and healthy controls group. This result is in line with (Bartosik et al., 2010) who also compared subjects with MTSS and healthy controls and found no difference in static foot posture. These findings are, however, in contrast with the findings of Yates and White (2004) who reported that subjects with a pronated foot type have a significantly higher risk of developing MTSS during basic naval training.

We were restricted to treadmill walking to be able to investigate 20 consecutive steps of kinematics and SEMG. However, mechanically there are no theoretical differences between overground walking and treadmill walking (vanIngenSchenau, 1980). Matsas et al. (2000) have shown that knee joint kinematics obtained from treadmill walking after 6 min of customization are not significantly different from those obtained from overground

walking. Further (Tulchin et al., 2010) compared multi-segment foot kinematics during treadmill and overground walking. Their study indicated that the difference between overground and treadmill walking in most parameters is smaller than the day-to-day variability in linear foot kinematics. However, the extracted parameters are not identical to ones used in this study, and further studies are needed to assess measures of complexity during treadmill and overground walking.

The SEMG time series in our study were time-normalized and linearly interpolated to 1000 points per stance phase of each stride. The interpolation may slightly change the shape of the signal. Moreover, the normalization of SEMG signal in time and in amplitude introduces some variance in the assessed estimators (Jackson et al., 2009; Madeleine et al., 2002) and may mask differences between healthy controls and patients with MTSS. Reliability of sample entropy derived from SEMG signals has only been investigated during muscle fatigue by Sung et al. (2010). They found entropy measures to be highly reliable and more reliable than spectral estimator like median frequency or slope of the median frequency. In this study, we preferred PE and PeSaEn to the more common approximate entropy (Pincus et al., 1991) and sample entropy (Richman and Moorman, 2000) due to their fast calculation and relative invariance. PE has recently been shown to be a valid estimate of time series complexity. Its simplicity, extremely fast calculation as well as its robustness and invariance to noise are important criteria when estimating time series complexity (Bandt and Pompe, 2002). PeSaEn was recently examined though a heuristic approach using permuted limited number of samples to estimate sample entropy (Samani and Madeleine, 2010). PeSaEn based on the analysis of estimation variability was found to be a fair estimate of time series complexity at much lower computational cost. More studies combining SEMG, kinetic and kinematics recordings are needed.

Conclusion

There is an inverse relationship between the complexity of foot kinematics and the complexity of the SEMG signal in patients suffering from MTSS. Patients with MTSS are characterized by higher complexity of SEMG signal of the tibialis anterior and soleus, but lower complexity in midfoot kinematics.

Acknowledgements

Foot and Ankle Research Northern Denmark. The Association of Danish Physiotherapists Research Fund. None of the sponsors had any influence on study.

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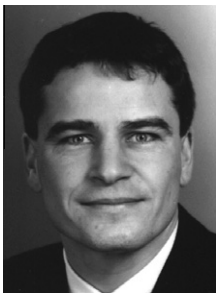

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