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Effects of wearing lower leg compression sleeves on locomotion economy

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ABSTRACT
The purpose of this investigation was to assess the effect of compression sleeves on muscle activation cost during locomotion. Twenty-two recreationally active men (age: 25 ± 3 years) ran on a treadmill at four different speeds (ordered sequence of 2.8, 3.3, 2.2, and 3.9 m/s). The tests were performed without (control situation, CON) and while wearing specially designed lower leg compression sleeves (SL). Myoelectric activity of five lower leg muscles (tibialis anterior, fibularis longus, lateral and medial head of gastrocnemius, and soleus) was captured using Surface EMG. To assess muscle activation cost, the cumulative muscle activity per distance travelled (CMAPD) of the CON and SL situations was determined. Repeated measures analyses of variance were performed separately for each muscle. The analyses revealed a reduced lower leg muscle activation cost with respect to test situation for SL for all muscles ($p < 0.05$, $\eta_p^2 > 0.18$). The respective significant reductions of CMAPD values during SL ranged between 4% and 16% and were largest at 2.8 m/s. The findings presented point towards an improved muscle activation cost while wearing lower leg compression sleeves during locomotion that have potential to postpone muscle fatigue.

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KEYWORDS
Ankle; SEMG; running; performance; efficiency

Introduction

Over the past two decades, elite athletes as well as recreationally active runners have been increasingly using compression garments to enhance their physical performance and to assist recovery. With this, a considerable body of literature has accumulated that is aimed at evaluating objectively the subjectively perceived enhanced performance and recovery (MacRae, Cotter, & Laing, 2011). Furthermore, basic investigations were performed to better understand the underlying mechanisms of compression sleeves (Born, Sperlich, & Holmberg, 2013). In principle, compressive garments are assumed to improve hemodynamics and therefore deeper tissue oxygenation (Agu, Baker, & Seifalian, 2004) by shifting superficial blood to the deeper vessels (Lawrence & Kakkar, 1980).

The increasing publicity in turn has had a positive impact of an even more frequent use of the respective clothing. Although there is some evidence that wearing compression clothing during exercise enhances maximal strength, explosive power and also endurance, the respective effect sizes were only small (Born et al., 2013). In addition to performance outcome measures, the effects of compression garments on, for example, energy expenditure using either direct or indirect measures was evaluated (Born et al., 2013; Bringard, Perrey, & Belluye, 2006; Dascombe, Hoare, Sear, Reaburn, & Scanlan, 2011). In contrast, the effects observed during recovery were much stronger (Bieuzen et al., 2014; Born et al., 2013).

According to Daniels (Daniels & Daniels, 1992) running economy is defined as the relationship between oxygen consumption and running velocity. The standard method to measure running economy is the steady state oxygen uptake (VO_2) at intensities below the ventilatory threshold (Foster & Lucia, 2007). VO_2 is typically measured in litres of oxygen per kilogram of body weight per minute. However, to express VO_2 (and other variables) as a function of distance travelled rather than as a function of time would help to compare the locomotion costs of subjects being tested at different running velocities (Vercruyssen, Gruet, Colson, Ehrstrom, & Brisswalter, 2017). Moreover, running economy comprises metabolic, cardiorespiratory, biomechanical and neuromuscular characteristics that are unique to the individual (Barnes & Kilding, 2015). Hence, changes of neuromuscular characteristics expressed as a function of distance travelled might offer a valuable alternative to compare muscle activation cost between two specific modes.

Interestingly, investigations that directly targeted the effects of compression garments on physiological muscle parameters are comparably rare. We could only find a small number of investigations utilizing Surface EMG (SEMG). These investigations showed reduced muscular effort with unchanged performance parameters (Fu, Liu, Zhang, Xiong, & Wei, 2012; Wang, Xia, & Fu, 2016), or even reduced signs of muscular fatigue (Miyamoto, Hirata, Mitsukawa, Yanai, & Kawakami, 2011; Zhang, Fu, Xia, & Wang, 2016). All these investigations analysed the aforementioned effects during or in relation to maximum effort, i.e. artificial test situations. Investigations that aimed at evaluating systematic effects of compression garments

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during approximated normal, i.e. submaximal test conditions, are lacking.

Therefore, this study was designed to identify the effect of lower leg compression sleeves on muscle activation cost during submaximal non-fatiguing locomotion, i.e. time- limited running on treadmill utilizing SEMG measurements.

Methods

Subjects

Twenty-two healthy and recreationally active male subjects were examined after having read and signed informed written consent forms. Their actual activity level on a one (no physical activity at all) to five (daily activity of more than one hour) activity scale was 4.0 (3.0/4.0, median and quartiles). Demographic data on the study participants are presented in Table 1. The study protocol was approved by the local ethics committee (4539–09/15) and therefore complied with the Declaration of Helsinki for human research.

Design

In this two period two task crossover study, each participant ran on a treadmill (Quasar.med, H-P-Cosmos) at four different speeds (ordered sequence of 2.8, 3.3, 2.2 and 3.9 m/s). Each of the tasks was completed once, in random order. There were eleven subjects for each order. Subjects visited the laboratory on two different occasions approximately three weeks apart. Participants begun either with normal, uncoated task (control condition, CON) or they were equipped with specially designed lower leg compression sleeves (sleeve condition, SL, Bauerfeind AG, see Figure 1, left section). All measurements were performed at the same time of the day for each participant under similar environmental conditions (room temperature ranged between 21 and 23°C). To avoid muscle fatigue, subjects did not run permanently, but instead rested between the single running trials (i.e. the non-stationary phases to adjust for the next speed) at the non-moving edges of the treadmill (see Figure 1, right section). Further, subjects only ran for approximately 30 strides per speed (start and stop announcements were provided).

Compression garment

As claimed by the manufacturer the lower leg compression sleeves worn (polyamide, elastane) had a decreasing pressure profile from distal (4.0 kPa) to proximal (2.7 kPa) to enhance venous return. The sizes of the compression sleeves ranged between 19 cm circumference above the malleoli (smallest size) and 51 cm the greatest circumference without compressing the subcutaneous tissue (largest size). The length of the lower leg segment (see Table 1) was measured as the distance between the circumference levels (shortest: 26 cm, longest: 43 cm) as was recommended by the manufacturer. Further, the lower leg sleeves had a specific working pressure which was generated by the shape change of the calf during movement and the selection of the material components in the individual zones.

Figure 1. Left section: Application of the compression sleeves. Right section: Subject running on the treadmill while wearing the sleeves. Please note the non-moving edges of the treadmill.

Methodology

SEMG of five superficial lower leg muscles (tibialis anterior (TA), fibularis longus (FL), lateral and medial head of gastrocnemius (LG, MG), and soleus (SO)) was simultaneously measured at both sides of the body. Electrode locations were chosen according to the SENIAM recommendations (Hermens et al., 1999) that also included adapted skin preparation (i.e. shaving, cleaning, and abrading). Disposable SEMG electrodes (H93SG, Covidien) with a circular uptake area (diameter: 1.6 cm) and an inter-electrode distance of 2.5 cm were applied. Signals were amplified (–3 dB at 5 Hz and 700 Hz, gain: 1,000, Biovision), with the amplifiers located close to the electrodes. The electrodes together with the amplifiers were carefully secured with highly elastic circular net bandages to avoid movement artefacts. Analog-to-digital conversion was carried out at a rate of 2,048/s (Tower of Measurement, amplitude resolution: 24 bit at ± 5 V, anti-aliasing filter at 1,024 Hz, DeMeTec). Raw data were stored on hard disk for subsequent offline analysis using the ATISArec capturing program (GJB). Pressure sensors were applied below both heels (FSR 402, Interlink) to detect strides unequivocally.

SEMG data were band-pass filtered between 20 Hz and 300 Hz. To account for interferences from the electrical current

Table 1. Characteristics of study participants.

	Age [years]	Height [cm]	Mass [kg]	BMI [kg/m^2]	Fat [%]	LL [cm]
Mean (SD)	25.3 (2.8)	182.7 (5.4)	77.5 (6.5)	23.2 (1.5)	14.7 (4.5)	32.9 (2.1)
Min, Max	22, 33	170, 190	66, 89	21, 27	5, 21	27, 36

BMI, Body Mass Index; LL, Lower Leg Segment Length.

supply, a 50 Hz notch filter was applied. Steady conditions were ensured by giving the subjects approximately three strides to adapt to the actual treadmill speed. For the analysis, only strides deviating less than 10% from the median time of all respective strides per subject and running speed were considered. Signals of the remaining strides were quantified as root mean square (rms) values and smoothed with a moving rectangular window of 50 ms. These amplitude curves were time-normalized on the basis of consecutive ipsilateral heel strikes (time resolution of 0.5%). Mean amplitude values were calculated by averaging the respective time-normalized amplitude curves. Muscle activation cost was determined by calculating the cumulative muscle activity per distance (CMAPD, (Carrier, Anders, & Schilling, 2011)) of the CON and SL situations, individually and separately for each muscle and speed. CMAPD was calculated by using the following equation:

$$CMAPD\left[\frac{\mu V \times s}{m}\right] = \frac{\frac{1}{n}\sum_{i=1}^{n} x_i}{v}$$

in which "x" are the time normalized values of the SEMG curve and "v" is the running speed.

Statistical analyses

No differences among sides were observed for CMAPD values. Thus, values of both sides were averaged and used for further analyses. Effects of applying lower leg compression sleeves on muscle activation cost (CMAPD) were analysed by comparing the CON and SL situations with repeated measures analyses of variance (speed (4) × sleeve (2)), separately for each muscle. In the event of significant interaction effects, results were visualized using profile plots according to Leigh and Kinnear (Leigh & Kinnear, 1980) to avoid interpretation errors. To account for different absolute values, the differences between CON and SL were analysed as relative values. One-way repeated measures analyses of variance followed by Fisher's Least Significant Difference (LSD) post hoc tests were utilized. Effect sizes were calculated as partial eta squared (η_p^2) with values ≥ 0.01, ≥ 0.06, ≥ 0.14 indicating small, moderate, and large effects respectively.

Table 2. P values of post hoc comparisons of speed-related influences (Fischer's LSD test).

speed [m/s]		TA	FL	LG	MG	SO
2.2	2.8	0.68	<0.001	<0.001	<0.001	<0.001
	3.3	0.29	0.24	0.083	0.93	<0.02
	3.9	0.74	0.076	<0.04	0.91	<0.001
2.8	2.2	0.68	<0.001	<0.001	<0.001	<0.001
	3.3	0.45	<0.002	<0.04	<0.002	<0.05
	3.9	0.93	<0.005	<0.03	<0.004	<0.03
3.3	2.2	0.29	0.24	0.083	0.93	<0.02
	2.8	0.45	<0.002	<0.04	<0.002	<0.05
	3.9	0.23	0.65	0.87	0.98	0.65
3.9	2.2	0.74	0.076	<0.04	0.91	<0.001
	2.8	0.93	<0.005	<0.03	<0.004	<0.03
	3.3	0.23	0.65	0.87	0.98	0.65

TA, tibialis anterior muscle; FL, fibularis longus muscle; LG, MG, lateral and medial head of gastrocnemius muscle; SO, soleus muscle.

Results

Application of lower leg compression sleeves reduced CMAPD levels for each muscle investigated ($p < 0.05$), with larger effects for the plantar flexors ($\eta_p^2 > 0.37$, see Table 2). The analyses further showed ordinal interactions between "speed" and "sleeve" for every muscle ($p < 0.01$, $\eta_p^2 > 0.21$, except for TA). Therefore, the main effects found could be interpreted independently of the interaction. In general, CMAPD values decreased in line with increasing running speed (Figure 2).

For all plantar flexors, the degree of the relative CMAPD reduction was subject to running speed ($F > 5.2$, $p < 0.01$, $\eta_p^2 > 0.2$), with the largest effect found for SO ($\eta_p^2 = 0.32$). The relative change of CMAPD values was always in favour for the SL situation, and for the verifiable reductions showed values between 4% and 16% (Figure 3). The results of the post hoc comparisons for the identification of speed-related influences proved the largest reductions at 2.8 m/s and are detailed in Table 2. However, for TA, no systematic influence of running speed on the relative CMAPD differences could be proven.

Discussion

This study showed clear beneficial effects of the applied compression sleeves on muscle activation cost during running. This was further subject to running speed and exhibited the largest reductions at 2.8 m/s.

In general, running economy is to be considered a complex conceptual framework that is subject to multiple lower-body attributes (Barnes, McGuigan, & Kilding, 2014). Within this framework, but from a more biomechanical perspective, shorter Achilles tendon moment arms correlated with better running economy (Barnes et al., 2014). Based purely on physiology, running economy can be determined through the steady-state oxygen consumption at a given running velocity (Daniels & Daniels, 1992). As was done in this study, the analysis of electrical energy required to activate a muscle at a given running velocity was performed. This was done to provide a neuromuscular basis for the known more general effects of compression garments.

With respect to effects specifically utilizing lower leg compression garments we could only find five studies (Bieuzen et al., 2014; Del Coso et al., 2014; Menetrier, Mourot, Bouhaddi, Regnard, & Tordi, 2011; Sambaher, Aboodarda, Silvey, Button, & Behm, 2016; Stickford, Chapman, Johnston, & Stager, 2015). Muscular performance (including countermovement jump) and physiological variables did not differ between a calf compression and a control group wearing regular athletic socks after a half-ironman triathlon race (Del Coso et al., 2014). In contrast, 48 h after finishing simulated trail races of 5.2 km, athletes recovered faster when they utilized leg compression, as revealed by countermovement jumping height (Bieuzen et al., 2014). Due to large inter-individual differences in responses to calf compression, no systematic effects regarding running economy or gait variables could be observed during running at three different speeds (Stickford et al., 2015). Although tissue oxygen saturation of the gastrocnemius muscle was found to be higher after exercise with calf compression, running performance related benefits could not be verified (Menetrier et al., 2011). Only one of the aforementioned studies that utilized lower leg compression sleeves

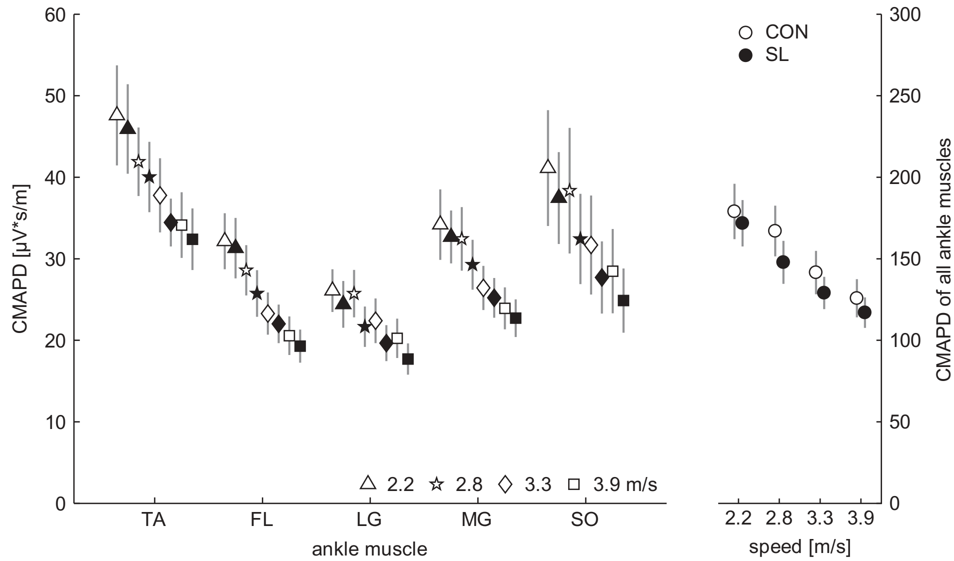

Figure 2. Absolute CMAPD values presented as mean and 95% confidence intervals.

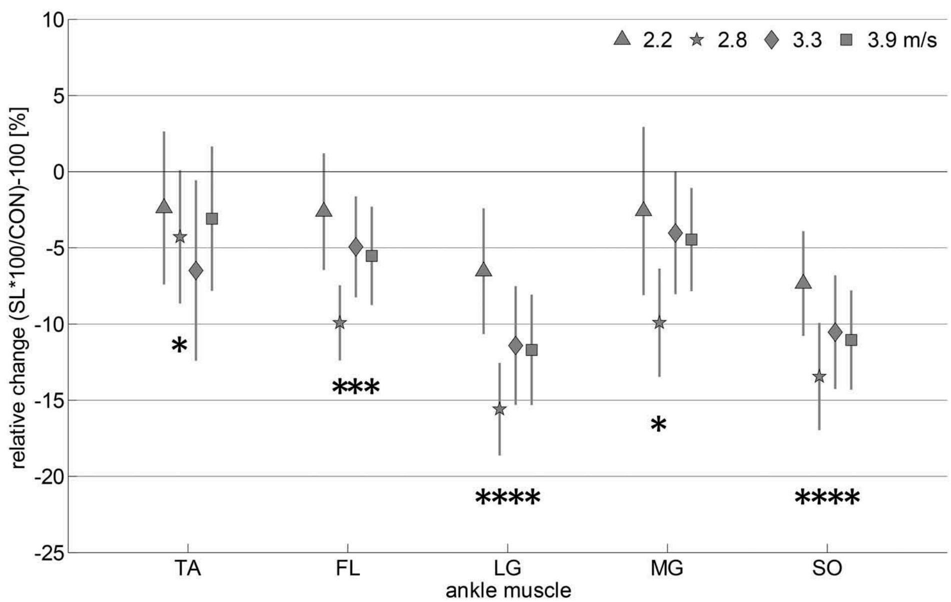

Figure 3. Relative changes of CMAPD values between CON and SL. Negative values indicate reduced CMAPD levels for SL. Values presented as mean and 95% confidence intervals. Asterisks indicate significant differences between CON and SL (t tests, Bonferroni correction included).

applied SEMG measurements on calf muscles. Contrary to the other studies, the participants were asked to perform a fatigue protocol of continuous drop jumps until task failure (Sambaher et al., 2016). One of the main outcomes was that the lower leg compression group revealed increased skin temperature as compared with controls. However, calf muscle SEMG did not differ between groups.

In young adults muscle activation explained 30% of variance in metabolic cost of level walking (Hortobagyi, Finch, Solnik, Rider, & DeVita, 2011). Further, in older individuals higher metabolic costs of walking were shown to be attributable to increased CMAPD of all lower limb muscles and explained up to 83% of variance (Pincheira, Stenroth, Avela, & Cronin, 2017). In the present work, the reduced muscle activation cost implies a positive effect of the compression garment on energy consumption at all speeds investigated. However, to equate muscle activation cost (CMAPD) during locomotor activities with metabolic cost of transport would oversimplify the situation. Thus, capturing muscle activity via surface EMG provides an indication of muscle metabolism (Blake & Wakeling, 2013; Pincheira et al., 2017) but is not able to measure muscle metabolism per se. Concerning the observed effect sizes on outcome variables our findings are in contrast with previously reported results on wearing compression garments (Engel, Holmberg, & Sperlich, 2016; Vercruyssen et al., 2017). Different to other commercially available compression garments studied (Vercruyssen et al., 2017) the design of the clothing applied in the present study might have induced a different effects, which probably reduced muscle activation costs. However, since the individual garment pressures were not obtained, no statements can be made about the actual pressure.

According to the distinct functions of the lower leg muscles investigated with respect to the ankle joints, we have investigated the triceps surae muscle (i.e. LG, MG, and SO) as isolated plantar flexor of the foot, the FL that in addition to plantar flexion also pronates and adducts the foot, and the TA that supinates and dorsally flexes the foot. Since compression garments are known to improve explosive power particularly (Born et al., 2013), we expected to see the most prominent effects for the muscles involved in plantar flexion. This could clearly be proven. Only the TA as a dorsal flexor showed inferior effects due to the compression sleeves. The TA enables the dorsiflexed forefoot position during early and mid-swing phase and eccentrically controls the foot posture during initial stance phase (Novacheck, 1998). Regarding effects of different speeds on muscle activation costs, our previous results have already demonstrated an energetic optimum at approximately 3.3 to 3.9 m/s for all lower leg muscles investigated (Carrier et al., 2011). Therefore, it is no surprise that the largest muscle strain reduction could be observed at the less efficient speed of 2.8 m/s, which might therefore benefit most from the compression sleeves. The slowest running speed of 2.2 m/s has to be considered as non-physiologically slow and might therefore not benefit sufficiently from the compression sleeves.

Previous investigations on performance enhancement through leg compression utilized mainly workloads with high to maximum intensities, and thus might have introduced signs of peripheral fatigue. By applying the current study protocol, fatiguing influences can be precluded with high certainty (i.e. only 30 complete strides per running speed and rest phases between steady speed conditions).

To summarize, the application of the lower leg compression sleeves systematically reduced the muscular effort of ankle muscles, but to different extents at different speeds and for different muscles. This improved muscle activation cost has the potential to increase muscle performance through a possible delay of fatigue, but provides only an indirect indication of benefit to performance and recovery. Saved muscular activity may thus also enhance active joint protection (i.e. stabilization), but at the same time bears the potential risk of overloading passive tissues.

However, this study has some limitations that need to be addressed: One major methodological limitation of this study was the use of a control condition which does not mimic the warming effect of the experimental condition. We cannot rule out the possibility that the reduced CMAPD values are at least partly caused by an elevated skin temperature, which is known to reduce SEMG amplitudes (Zipp, 1977). Merletti, Sabbahi, & De Luca (1984) found a relationship between the initial median frequency of the myoelectric signal and reduction of intramuscular temperature. Surface cooling led to significant reductions of the initial median frequency. However, in their study Merletti et al. (1984) did not reported amplitude estimations. As was shown for endurance contractions without force declines (Arendt-Nielsen & Mills, 1988) and reduced temperature above the calf (Winkel & Jorgensen, 1991) the SEMG signal was accompanied with increased amplitude values. Moreover, the compression garment likely modified the distance to the muscle. A lower cross-section of the subcutaneous tissue would result in a signal that would be less low-pass filtered and thus dampened (Kuiken, Lowery, & Stoykov, 2003). Thus, signals with higher amplitudes would be measured on the skin surface. Therefore, effects of warming and compression through the garment on the SEMG amplitudes seem negligible. Further, individual garment pressures were not included.

Conclusion

Application of lower leg compression during treadmill running for short distances reduced CMAPD levels independent of the ankle muscle investigated. The findings presented point towards an optimized muscle activation cost of lower leg muscles while wearing compression sleeves during sub-maximal dynamic effort that has the ability to postpone muscle fatigue during endurance locomotion.

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Disclosure statement

No potential conflict of interest was reported by the authors.

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