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Tibial impacts and muscle activation during walking, jogging and running when performed overground, and on motorised and non-motorised treadmills

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Highlights

- Non-motorised treadmill locomotion creates large reductions in tibial acceleration.
- Non-motorised treadmill locomotion increases lower limb muscular activation.
- Non-motorised treadmill locomotion decreases cycle time/increases step frequency.

Abstract

Purpose: To examine tibial acceleration and muscle activation during overground (OG), motorised treadmill (MT) and non-motorised treadmill conditions (NMT) when walking, jogging and running at matched velocities.

Methods: An accelerometer recorded acceleration at the mid-tibia and surface EMG electrodes recorded rectus femoris (RF), semitendinosus (ST), tibialis anterior (TA) and soleus (SL) muscle activation during OG, MT and NMT locomotion whilst walking, jogging and running.

Results: The NMT produced large reductions in tibial acceleration when compared with OG and MT conditions across walking, jogging and running conditions. RF EMG was small-moderately higher in the NMT condition when compared with the OG and MT conditions across walking, jogging and running conditions. ST EMG showed large and very large increases in the NMT when compared to OG and MT conditions during walking whilst SL EMG found large increases on the NMT when compared to OG and MT conditions during running. The NMT condition generated very large increases in step frequency when compared to OG and MT conditions during walking, with large and very large decreases during jogging and very large decreases during running.

Conclusions: The NMT generates large reductions in tibial acceleration, moderate to very large increases in muscular activation and large to very large decreases in cycle time when compared to OG and MT locomotion. Whilst this may decrease the osteogenic potential of NMT locomotion, there may be uses for NMTs during rehabilitation for lower limb injuries.

Key Words: Accelerometer, EMG, Biomechanics, Treadmill, Locomotion

Introduction

Walking and running are the most common forms of human locomotion and are usually performed overground. However, walking and running are often performed on treadmills as attractive alternatives and to facilitate studies under controlled conditions.

The motorised treadmill is the most common ergometer and is powered by a motor that keeps the treadmill belt at a constant velocity. The non-motorised treadmill is less common and is characterised by a freely moveable treadmill belt powered by the individual by means of a horizontal tether attached at the waist. This allows the self-propelled belt to rotate according to the speed of the participant. Several studies have compared motorised treadmill vs overground locomotion to examine kinematics [1], ground reaction forces [2] and muscular activation differences [3,4,5]. Similarly, non-motorised treadmill and overground locomotion have been compared for 5000 m performance time, electromyography (EMG), blood lactate, oxygen uptake kinetics, heart rate [6], maximal sprinting performance [7] and 6-minute walk distance [8] that have all highlighted dissimilarities between the conditions which could affect the mechanical loading environment and also the musculoskeletal adaptations generated by different locomotion conditions.

Walking and running, either overground or on a treadmill are recommended for the health of the general population [9], with benefits including reduced body fat, lowered resting heart rate and increased maximal oxygen uptake [10]. Walking and running are also recommended for maintaining bone health during ageing [11,12,13,14]. For bone health, it is important to establish the magnitude of mechanical loading and muscle activation generated by walking and running as the intensity of loading encourages skeletal adaptation [15]. Muscular activation has been linked with internal compressive forces that increase the mechanical loading on bones [16]. In addition, muscles impose a stress on the skeletal system which increases bone remodelling [17]. Impact forces and muscle activation patterns are well recognised in the habitual human gait, with accelerometry and EMG showing the forces experienced and internal muscle activity [18,19,20]. Due to the biomechanical differences between overground, motorised treadmill and non-motorised treadmill conditions, there is also the potential for the impact forces and EMG to show differences across the locomotion conditions which would alter the mechanical loading environment. It is therefore important to establish the mechanical loading generated during each condition to determine their osteogenic potential.

Given the popularity of walking and running overground and on treadmills, it is important to understand how the impacts and muscle activity respond under different conditions in these types of locomotion. We hypothesise that as differences have been highlighted in a number of physiological variables during NMT locomotion compared with overground locomotion, that the impact forces and EMG may also be altered when using a NMT which could change the mechanical loading stimulus for musculoskeletal adaptations. This is the first study to comprehensively examine impacts and muscle activation during locomotion at different velocities and in different conditions within the same population. Accordingly, the aim of this study was to examine the ground impacts via accelerometry (ACC) and muscle activation via surface EMG generated during overground (OG), motorised treadmill (MT) and non-motorised treadmill (NMT) conditions when walking, jogging and running at matched speeds.

Methods

Participants

All 15 participants (mean ± SD: 24.2 ± 3.8 y, 179.5 ± 3.9 cm, 81.0 ± 7.2 kg) were recreationally active. Familiarisation was undertaken at least 48 h before the main testing, and involved walking, jogging and running at a constant speed on a non-motorised treadmill (NMT). Participants were already familiar with overground and motorised treadmill locomotion. The protocol was approved by the institutional ethics committee and informed consent was obtained from all participants prior to testing.

Procedures

Tibial acceleration and lower body muscle activation were measured during OG, MT and NMT locomotion whilst walking, jogging and running at matched velocities using a cross sectional repeated measures design. Following a warm up of walking, jogging, running and dynamic stretching, participants walked, jogged and ran along a 40 m indoor laboratory at a self-selected constant velocity whilst instantaneous velocity was recorded at 100 Hz with a speed meter via a waist harness (Speed Real Time, AP Lab, V3.1 – 2012, Rome, Italy). Trials were repeated if necessary to achieve a constant velocity (determined from manual inspection of velocity data). Overground walking (1.56 ± 0.15 $m \cdot s^{-1}$), jogging (2.88 ± 0.35 $m \cdot s^{-1}$) and running (4.28 ± 0.36 $m \cdot s^{-1}$) were individually replicated during 30 s bouts on a MT (Woodway ELG55, Woodway, Weil an Rhein, Germany) and NMT (Woodway Force 2.0. Woodway, Weil an Rhein, Germany) in a randomised order. MT speeds were constant whereas NMT speeds were matched when walking (1.56 ± 0.13 $m \cdot s^{-1}$), jogging (2.88 ± 0.35 $m \cdot s^{-1}$) and running (4.25 ± 0.37 $m \cdot s^{-1}$). Participants were instructed to walk, jog or run “naturally”. Trials were separated by 4-5 min rest allowing sufficient recovery and to reduce any effects of fatigue. Umbro 5v5 trainers (Umbro, Cheshire, UK) were worn by all participants in their correct size to standardise footwear.

ACC and EMG data were collected synchronously (sampling rate = 1500 Hz, input impedance > 100 MΩ, CMRR > 100 dB, baseline noise < 1μV RMS, base gain = 200, final gain = 500) and stored on a computer using a 16-bit resolution wireless system (Desktop DTS, Noraxon USA Inc, Arizona, USA). An accelerometer (DTS 3D accelerometer-16 g, Noraxon USA Inc, Arizona, USA) was attached to the mid-anterior right tibia (50% of the distance between the tibial tuberosity and medial malleolus). Surface EMG electrodes (Ambu Blue Sensor N, Ambu, Cambridgeshire, UK) were placed over the rectus femoris (RF), semitendinosus (ST), tibialis anterior (TA), and soleus (SL) muscles of the participant's right leg in accordance with SENIAM surface electromyography recommendations [21]. Prior to electrode attachment, the skin was shaved, abraded and cleansed with a 70% alcohol swab. ACC and EMG wearable hardware were secured with surgical tape and elasticated bandages to reduce unwanted movement and signal artefacts.

Data Processing

Each gait cycle was identified using tibia accelerometer data, beginning at the lowest trough preceding the impact peak of the right tibia (which represented initial ground contact) and ending at the same point preceding the next impact peak of the right tibia [22]. Eight cycles were selected for analysis from a section where the participant was moving at a matched constant velocity in each condition. Point of ground contact was established using pilot data where synchronised motion capture, ground reaction force, sacrum and tibia accelerometers were used.

ACC data was low-pass filtered at cut-offs of 16, 33 and 40 Hz for walking, jogging and running respectively across all conditions based on a cut-off frequency set at 95% of the signal energy from a mean of the trials from the first 10 participants [23]. Acceleration peak was established as the immediate impact peak following ground contact. Acceleration gradient was calculated as the slope from the point of ground contact to the acceleration peak [24] and cycle time was calculated as the duration between right foot ground contacts upon landing. Acceleration peak, acceleration gradient and cycle time were averaged across 8 cycles per trial.

EMG data was band-pass filtered (bi-directional Butterworth, 10-500 Hz), full wave rectified and low-pass filtered at 15 Hz to obtain linear envelopes. EMG amplitude was calculated as the area under the curve (trapezium method) for each of the 8 identified cycles. EMG amplitude was taken as the mean across 8 cycles per trial and normalised to the NMT run trial. EMG co-contraction values were calculated, expressing the EMG amplitude of the agonist musculature as a percentage of the antagonistic musculature. RF values were expressed as a percentage of the ST values whilst TA values were expressed as a percentage of the SL values. A value of 100 indicates equal activation of the agonist and antagonist muscles. Values over 100 indicate greater RF or greater TA muscle activation compared to the ST and SL muscles respectively [25].

Data were processed using Myoresearch XP software (Myoresearch XP Master Edition 1.08.27, Noraxon USA Inc, Arizona, USA) and a bespoke MATLAB programme (MATLAB R2011a, Mathworks, Cambridge, UK).

Statistical Analysis

Data containing excessive signal interference were removed. Parametric data were statistically analysed using two-way (3 conditions x 3 velocities) repeated measures ANOVAs (Sidak adjustments) with post-hoc pairwise comparisons using SPSS (IBM SPSS Statistics Version 20.0. IBM Corp, NY, USA). Where applicable, non-normally distributed data were log-transformed and analysed using parametric methods. Cohen's *d* effect size is reported and evaluated using the following scale: 0-0.19 trivial, 0.2-0.59 small, 0.6-1.19 moderate, 1.2-1.99 large, 2.0-3.99 very large. Uncertainty in the population estimates are expressed as 95% confidence intervals along with the likelihood that the effect is substantially positive, trivial or substantially negative [26]. For non-parametric data where log-transformation was not possible, Friedman's tests were used to compare main effects of treadmill and velocity conditions. Wilcoxon signed-rank tests with Bonferroni corrections were used for pairwise comparisons and post hoc tests in SPSS, resulting in an alpha level set at $P < 0.017$ due to 3 groupings. Cliff's Delta (δ) effect size was calculated in R (R Foundation for Statistical Computing 3.2.1, Vienna, Austria) and evaluated using the following scale: 0-0.146 trivial, 0.147-0.32 small, 0.33-0.473 moderate, >0.474 large. Uncertainty in the population estimates are expressed as 95% confidence intervals [27].

Results

Accelerometry

The running condition resulted in large increases in acceleration peaks and gradients when compared to walking and jogging trials, while jogging trials resulted in large increases in acceleration peaks and gradients when compared to walking trials (Fig. 1). The NMT produced large reductions in acceleration peaks when compared to OG and MT conditions across all walking (δ = -0.56 [95%CI: -0.81 to -0.13], P = 0.004; , δ = -0.58 [95%CI: -0.83 to -0.15], P = 0.002), jogging (δ = -0.64 [95%CI: -0.85 to -0.23], P = 0.001; δ = -0.78 [95%CI: -0.92 to -0.45], P = 0.001) and running conditions (δ = -0.51 [95%CI: -0.77 to -0.11], P = 0.004; δ = -0.51 [95%CI: -0.78 to -0.01], P = 0.001). OG and MT conditions were similar. The treadmill condition had no effect on acceleration gradients across walking (P = 0.931), jogging (P = 0.155) and running (P = 0.395) (Fig. 2).

EMG Amplitude

There was a small to very large increase in all EMG amplitudes (four muscles) during the running condition when compared to walking and jogging trials, while jogging trials generated small to large increases in all EMG amplitudes above walking trials (Table 1). There was a small increase in RF EMG during the NMT condition when compared to the OG condition and a moderate increase when compared to the MT condition across all gait conditions (Table 1). The NMT generated large increases in ST EMG when compared to the OG condition and very large increases in ST EMG when compared to the MT condition whilst the OG condition gave small increases in ST EMG when compared to the MT condition during walking (Table 1). ST EMG was similar across all conditions for jogging and running. OG and NMT conditions produced moderate increases in TA EMG when compared to the MT condition during walking, additionally the OG condition produced moderate and small increases in TA EMG when compared to the MT condition for jogging and running respectively (Table 1). OG and NMT conditions created large increases in SL EMG when compared to the MT condition during jogging, while the NMT created large increases in SL EMG when compared to OG and MT conditions during running (Table 1).

Co-contraction RF/ST

No statistical differences were observed between conditions during walking trials, however the NMT condition generated moderate increases in co-contraction values when compared to OG conditions and small increases when compared to MT conditions during jogging (d = 0.69 [95%CI: 0.17 to 1.21], P = 0.033; d = 0.33 [95%CI: -0.42 to 1.09], P = 0.006). The NMT condition generated very large increases in co-contraction values when compared to OG and MT conditions during running (d = 2.57 [95%CI: 1.05 to 4.1], P = 0.024; d = 2.2 [95%CI: 0.93 to 3.47], P = 0.038) (Fig. 3).

Co-contraction TA/SL

No statistical differences were observed between conditions during walking trials, however the NMT condition displayed small reductions in co-contraction values when compared to OG and MT conditions during jogging (d = -0.4 [95%CI: -0.6 to -0.2], P = 0.009; d = -0.35 [95%CI: -0.61 to -0.1], P = 0.027) and running trials (d = -0.44[95%CI: -0.71 to -0.17], P = 0.023; d = -0.45 [95%CI: -0.75 to -0.14], P = 0.028) (Fig. 4).

Cycle Time

The NMT condition generated very large decreases in cycle time when compared to OG and MT conditions during walking (d = -2.24 [95%CI: -2.85 to -1.63], P < 0.001; d = -2.04 [95%CI: -2.64 to -1.44], P < 0.001), with large and very large decreases during jogging (d = -1.98 [95%CI: -2.5 to -1.46], P < 0.001; d = -2.03 [95%CI: -2.56 to -1.51], P < 0.001) and very large decreases during running (d = -3.03 [95%CI: -3.73 to -2.33], P < 0.001; d = -2.55[95%CI: -3.24 to -1.86], P < 0.001). OG and MT conditions produced similar cycle times.

Discussion

The main findings were that exercising on a NMT resulted in large reductions in peak acceleration on impact across all walking, jogging and running trials when compared to OG and MT conditions. Additionally, the NMT condition generated small and moderate increases in RF EMG (all trials), large and very large increases in ST EMG (walking) and large increases in SL EMG (running) in comparison with OG and MT conditions respectively. Findings indicate that habitual locomotion is altered when using a NMT which may decrease the level of mechanical loading and potentially the osteogenic nature of the exercise (as determined by the OI) but could provide useful rehabilitation purposes due to the reduction in impact forces.

Differences in ground reaction forces have been reported between OG and MT locomotion with the OG condition generating a 6% higher bodyweight percentage [2], although our data indicates that OG and MT conditions give similar acceleration peaks which supports more recent research [28]. The large reduction in peak acceleration during NMT locomotion could be caused by a pronounced forward lean favouring forefoot striking as opposed to heel/midfoot striking, but this assertion warrants research using motion analysis [29]. Large reductions in acceleration peaks during NMT conditions suggest that it is unsuitable for eliciting an osteogenic response, as previously determined thresholds (>4.9 g accounting for standing being 1 g) required to stimulate an increase in bone remodelling [15] are not consistently met, while OG and MT conditions elicit acceleration peaks above the required threshold when running (Fig. 1). Peak acceleration has been shown to have a graded effect on BMD adaptations with higher accelerations eliciting greater bone adaptations [15]. In the current study, this would indicate that the NMT could reduce BMD adaptations when compared to OG and MT conditions. Peak accelerations showed good agreement with previous MT studies [20,29] and the large reductions in acceleration peaks during the NMT condition are meaningful in surpassing 0.17 g, which has been suggested to be the minimal detectable change when walking [18]. This might imply that the NMT is better for rehabilitation when gradually re-introducing impact activity to individuals with lower extremity injuries. Acceleration gradient is also important in stimulating bone adaptation with OG, MT and NMT conditions producing similar results for this variable. Jogging and running consistently surpassed the threshold of 1000 $m \cdot s^{-3}$ indicating that adaptations due to this variable might be similar [24]. Acceleration gradients remained similar across treadmill conditions whereas acceleration peaks showed a large reduction during the NMT condition, this may have been caused by a shift in NMT kinematics which preserved the gradient of the acceleration on ground contact but cushioned the magnitude of the acceleration.

The NMT condition produced small and moderate increases in RF EMG when compared to all OG and MT trials, with large and very large increases in ST activation during walking and large increases in SL EMG during running. While the NMT could enhance training adaptations in these muscle groups it will alter natural OG movement patterns which highlights that information gathered from NMT locomotion should be interpreted with caution. Increased RF EMG has been linked with high internal compressive forces on the femur [16] with muscle action suggested as the main driver of bone adaptation [17]. Despite generating large reductions in peak acceleration, NMT locomotion could potentially initiate skeletal adaptations at the hip, which for osteoporotic patients could provide a means of stimulating bone maintenance without risk of osteoporotic fractures from high impact activity. This warrants further investigation as debate over the main stressor for bone adaptation continues and data from EMG studies are yet to show causal evidence for bone adaptation [17,30]. No statistical difference was observed between RF EMG in the OG and MT (-1% lower) conditions. This is contrary to previous studies, where up to 130% larger RF EMG values were reported during the MT condition while walking [3,4,5]. These differences are possibly due to walkway limitations (< 8 m), an elliptical walkway or inconsistent footwear.

The moderate and small increases in RF/ST co-contraction data during the NMT condition while jogging and the very large increase in RF/ST co-contraction data while running (in comparison with OG and MT conditions) indicates that a proportionally higher RF input was present. The small reductions in TA/SL co-contraction data during the NMT condition implied that a higher SL contribution was present throughout jogging and running trials. OG and MT trials were similar but the NMT appeared to induce higher levels of RF and SL stress with the exception of walking trials, which indicates that different jogging and running techniques are required to match OG velocities. This suggests that the NMT creates large to very large reductions in cycle times and therefore increases step frequency in order to match OG and MT velocities. This would also question the similarity of NMT locomotion and OG locomotion.

Limitations

Direction of force transfer is difficult to ascertain with skin mounted accelerometers due to orientation of the tibia, although few alternatives permit acceleration recording over multiple cycles during overground locomotion [18]. One familiarisation session on the NMT may be insufficient, two familiarisations could be optimal [31]. However, previous studies did not use only constant velocities which might negate the need for extra familiarisation, particularly as one familiarisation session has shown good reliability [32] and that participants all sufficiently met target velocities during familiarisation. High inter and intra participant variability for EMG was present, likely due to individual walking, jogging and running techniques which was expected, yet unavoidable.

Conclusion

In summary, the NMT generates large reductions in tibial acceleration, large to very large increases in step frequency and small to very large increases in muscular activation when compared to OG and MT locomotion. The reduction in tibial accelerations during NMT locomotion might reduce osteogenic adaptation, although could better suit individuals avoiding high impact exercise due to ongoing rehabilitation for lower limb injuries. The greater EMG response to NMT locomotion could indicate a higher training stimulus and higher internal compressive forces on the skeletal system which has been suggested to create a larger osteogenic stimulus, although this would require further investigation due to insufficient causal evidence for higher EMG and higher bone remodelling rates.

Conflict of interest statement

The authors report that there are no conflicts of interests.

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Figure 1

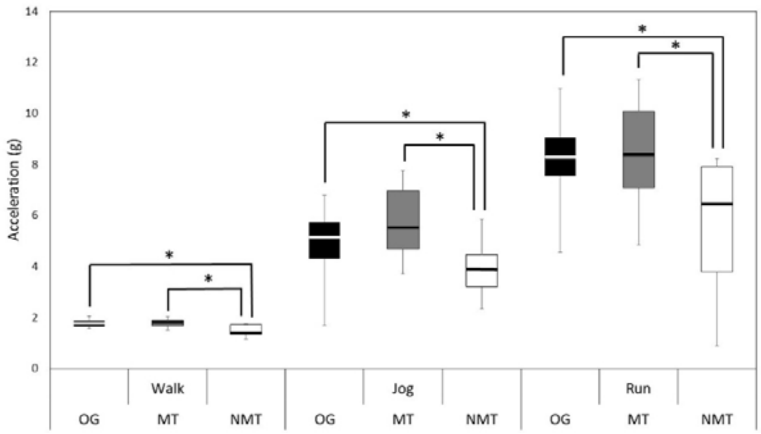


Figure 2

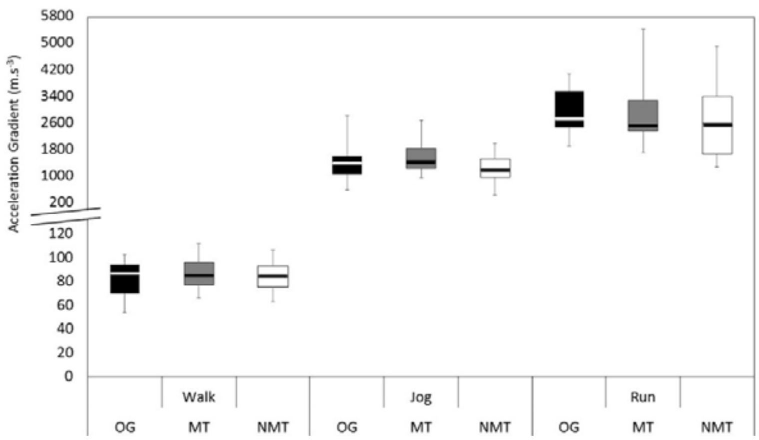
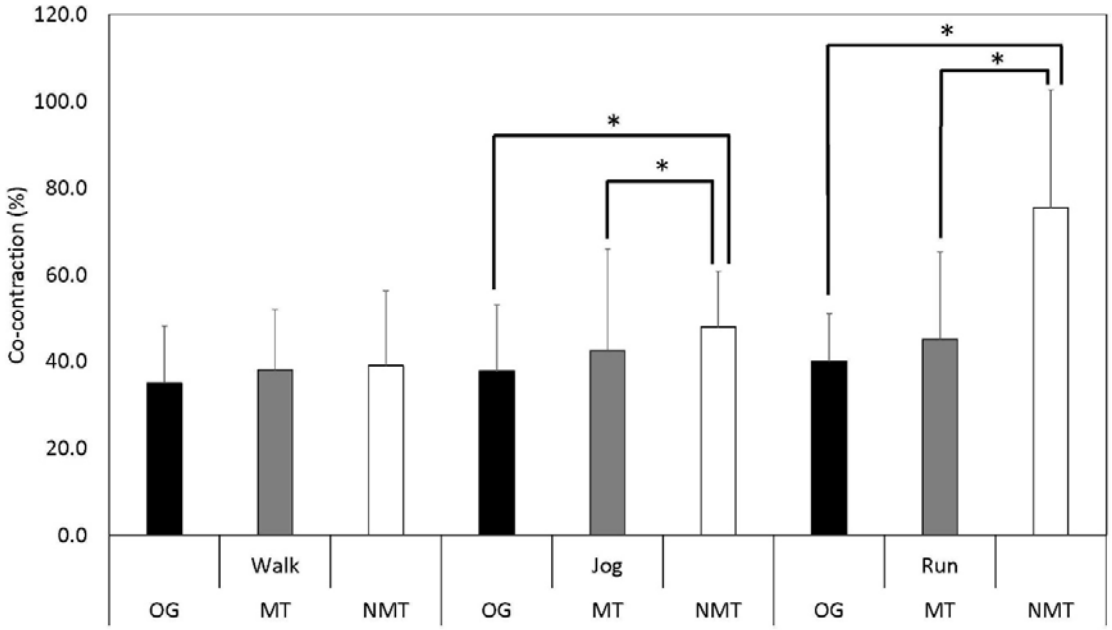


Figure 3



120.0
100.0
80.0
60.0
40.0
20.0
0.0
Co-contraction (%)
*
*
*
*
Walk
Jog
Run
OG
MT
NMT
OG
MT
NMT
OG
MT
NMT

Figure 4

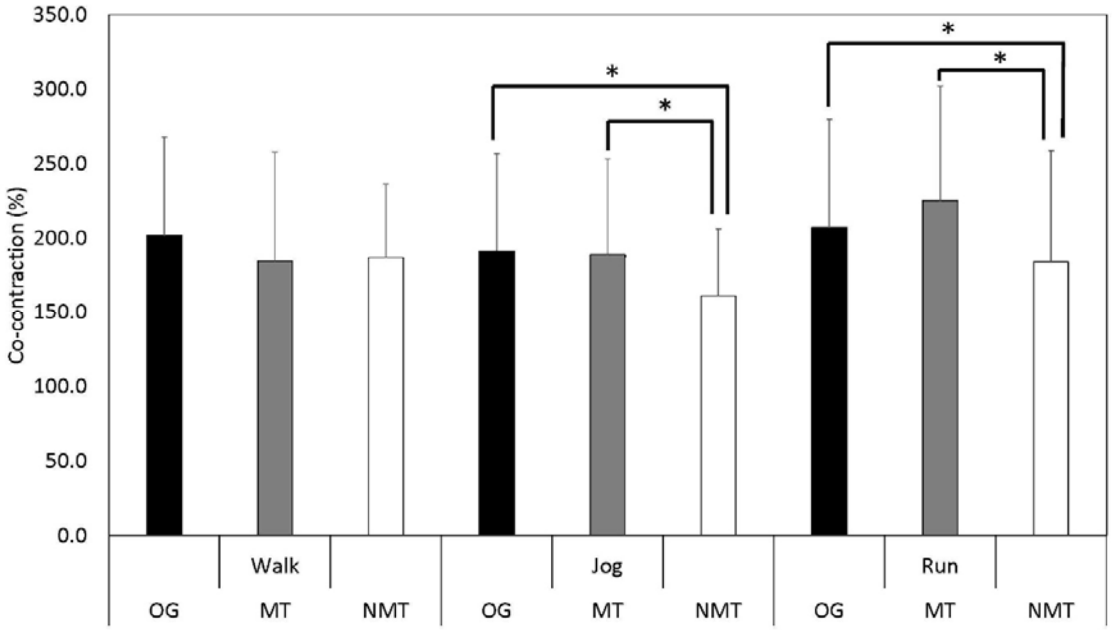

350.0
300.0
250.0
200.0
150.0
100.0
50.0
0.0
Co-contraction (%)
*
*
*
*
Walk
Jog
Run
OG
MT
NMT
OG
MT
NMT
OG
MT
NMT

Table 1

EMG amplitude (area under the curve) for each of the four muscles across overground, motorised treadmill and non-motorised treadmill conditions whilst walking, jogging and running. EMG amplitude is normalised to the NMT running trial and presented as a percentage.

	EMG Amplitude (Mean±SD)			***p* value**			**Cohen's *d* [95% confidence intervals]**		
RF	**OG**	**MT**	**NMT**	**OG-MT**	**OG-NMT**	**MT-NMT**	**OG-MT**	**OG-NMT**	**MT-NMT**
Walking	22 ± 12	22 ± 10	35 ± 17	*p*=0.997	*p*=0.015	*p*=0.015	*d* = 0.06 [-0.15 to 0.28]; 90% trivial	*d* = -1.06 [-1.65 to -0.48]; 100% negative	*d* = -0.96 [-1.65 to -0.27]; 98% negative
Jogging	38 ± 16	40 ± 18	52 ± 17	*p*=0.628	*p*=0.012	*p*=0.012	*d* = -0.14 [-0.42 to 0.15]; 67% trivial	*d* = -0.86 [-1.34 to -0.38]; 99% negative	*d* = -0.57 [-1.00 to -0.14]; 96% negative
Running	70 ± 31	69 ± 20	100 ± 0	*p*=0.993	*p*=0.02	*p*<0.001	*d* = 0.07 [-0.25 to 0.40]; 75% trivial	*d* = -0.97 [-1.54 to -0.41]; 99% negative	*d* = -0.95 [-1.39 to -0.51]; 100% negative
ST	**OG**	**MT**	**NMT**						
Walking	39 ± 10	36 ± 9	58 ± 15	*p*=0.04	*p*<0.001	*p*<0.001	*d* = 0.36 [0.09 to 0.63]; 88% positive	*d*= -1.81 [-2.46 to -1.17]; 100% negative	*d* = -2.17 [-2.91 to -1.43]; 100% negative
Jogging	64 ± 11	65 ± 11	68 ± 9	*p*=0.965	*p*=0.34	*p*=0.605	*d* = -0.07 [-0.40 to 0.27]; 74% trivial	*d* = -0.32 [-0.75 to 0.11]; 72% negative	*d* = -0.26 [-0.73 to 0.22]; 60% negative
Running	96 ± 19	89 ± 19	100 ± 0	*p*=0.538	*p*=0.843	*p*=0.125	*d* = 0.34 [-0.24 to 0.93]; 70% positive	*d* = -0.19 [-0.71 to 0.34]; 48% negative	*d* = -0.53 [-1.04 to -0.02]; 91% negative
TA	**OG**	**MT**	**NMT**						
Walking	83 ± 24	68 ± 18	89 ± 22	*p*=0.008	*p*=0.131	*p*=0.002	*d* = 0.61 [0.25 to 0.96]; 99% positive	*d* = -0.24 [-0.47 to -0.01]; 64% negative	*d* = -0.86 [-1.26 to -0.43]; 100% negative
Jogging	92 ± 21	74 ± 20	81 ± 18	*p*=0.003	*p*=0.185	*p*=0.142	*d* = 0.79 [0.38 to 1.19]; 100% positive	*d* = 0.48 [-0.04 to 1.00]; 87% positive	*d* = -0.31 [-0.61 to 0.00]; 76% negative
Running	111 ± 31	95 ± 28	100 ± 0	*p*=0.039	*p*=0.461	*p*=0.857	*d* = 0.50 [0.12 to 0.88]; 94% positive	*d* = 0.34 [-0.18 to 0.86]; 71% positive	*d* = -0.16 [-0.63 to 0.31]; 51% trivial
SL	**OG**	**MT**	**NMT**	**Bonferroni Correction (p<0.017)**			**Cliff's Delta (δ) [95% confidence intervals]**		
Walking	73 ± 29	61 ± 17	90 ± 30	*p*=0.041	*p*=0.002	*p*=0.012	δ = 0.12 [-0.28 to 0.49]	δ = -0.2 [-0.55 to 0.22]	δ = -0.37 [-0.69 to 0.08]
Jogging	81 ± 25	73 ± 14	91 ± 14	*p*=0.001	*p*=0.609	*p*=0.001	δ = 0.49 [0.04 to 0.78]	δ = -0.11 [-0.50 to 0.32]	δ = -0.65 [-0.87 to -0.21]
Running	91 ± 16	77 ± 11	100 ± 0	*p*=0.001	*p*=0.056	*p*=0.016	δ = 0.33 [-0.06 to 0.64]	δ = -0.57 [-0.85 to -0.04]	δ = -0.71 [-0.92 to -0.19]

* Soleus Data are presented as Medians ± Interquartile Range

RF, rectus femoris, n=12; ST, semitendinosus, n=15; TA, tibialis anterior, n=15; SL, soleus, n=13;

OG, overground; MT, motorised treadmill; NMT, non-motorised treadmill