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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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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.

1. 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 [3–5]. 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-min 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 [11–14]. For bone health, it is important to establish the magnitude

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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 [18–20]. 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.

2. Methods

2.1. 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.

2.2. 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 ($1.56 \pm 0.15\ \mathrm{m\,s^{-1}}$), jogging ($2.88 \pm 0.35\ \mathrm{m\,s^{-1}}$) and running ($4.28 \pm 0.36\ \mathrm{m\,s^{-1}}$) 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 ($1.56 \pm 0.13\ \mathrm{m\,s^{-1}}$), jogging ($2.88 \pm 0.35\ \mathrm{m\,s^{-1}}$) and running ($4.25 \pm 0.37\ \mathrm{m\,s^{-1}}$). 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 < 1 μV 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.

2.3. 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).

2.4. Statistical Analysis

Data containing excessive signal interference were removed. Parametric data were statistically analysed using two-way (3