GAIT KINEMATICS AND VARIABILITY DURING NORMAL AND UNWEIGHTED TREADMILL RUNNING

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The biomechanics of running in unusual environments has received little attention. Hence the aim was to compare normal to unweighted at 40% of bodyweight treadmill running. Kinematic data were recorded for six participants running at 3.35 m/s. The stride frequency during unweighted was less than normal treadmill running (67 ± 10 v 83 ± 5 strides/min; p = 0.03). Particularly during the stance phase, participants demonstrated less knee flexion-extension and ankle dorsi-plantar flexion angular motion. Participants also demonstrated greater within- and between-participant variability during unweighted than normal treadmill running (e.g. left knee angle SD of 1.7 ± 0.2° v 3.1 ± 2.1°; p = 0.03). Unweighted treadmill running produces gait kinematics with less movement and with greater variability, which may have implications for training and rehabilitative uses.

KEY WORDS: biomechanics, gravity, sport, technology.

INTRODUCTION: Human motion is performed in different environments ranging from air to water and low to high gravity conditions. In comparing overground to in-water running, significant coordination profile differences were found, such as, maximal knee extension overground of 59 ± 10° versus in-water of 39 ± 10° (Kilding et al., 2007). In using in-water running for rehabilitation, in contrast to the benefits, the authors proposed that consideration should be made of whether different muscle recruitment could further expose individuals to further injury or influence the transfer of technique back to overground running. In investigating motion in simulated reduced gravity, the Zero Gravity Locomotion Simulator (ZGLS) has been used in which participants run by being suspended supine from the ceiling and 'pulled' towards a vertical treadmill (McCrory et al., 2004). Compared to overground running, peak forces were less, loading rates higher and kinematics altered including greater knee flexion at heel-strike of 19 ± 2° during 'waist and shoulder springs' ZGLS running versus 7 ± 3° overground.

Another method of simulating reduced gravity has involved the AlterG treadmill where participants, sealed at the waist into an airtight enclosure encompassing a treadmill, have reduced weight by the air pressure inside the enclosure being increased to provide a counterforce to gravity. With less body weight, decreases in impact peaks, active peaks, vertical loading rates, gross metabolic power and stride frequency, and increased stance time have been observed (Grabowski and Kram, 2008). At 3 m/s the stance times were 0.2 ± 0.01 s (100%BW) and 0.25 ± 0.01 s (28%BW), and the stride frequencies were 1.49 ± 0.05 Hz (100%BW) and 1.32 ± 0.03 Hz (28%BW). With regard to the AlterG treadmill, the effects of unweighting on a person's running kinematics have not been investigated. Knowledge of this would have implications for specific training situations, such as motion in reduced gravitational environments, or more commonly on Earth as a rehabilitation tool. In this first instance the aim of this study was to compare the kinematics and variability of healthy participants treadmill running during normal and unweighted conditions.

METHODS: Six healthy recreational runners volunteered and provided written informed consent to participate in the study (height = 1.72 ± 0.09 m; mass = 74 ± 15 kg). All procedures were approved by the institution's ethics review committee. The participants wore their normal athletic shoes, close fitting shorts and no tops for males and sports-tops for females. A modified Cleveland Clinic marker set, including four-marker rigid-shell clusters on the shanks and thighs (Cappozzo et al., 1997), comprising of 80 retroreflective spherical markers between 5-10 mm in diameter were placed on the participants using adhesive tape (Figure 1).

8 cameras positioned behind and to the side, and above and below the treadmill surface, were used to see the upper body, thigh, shank and foot markers
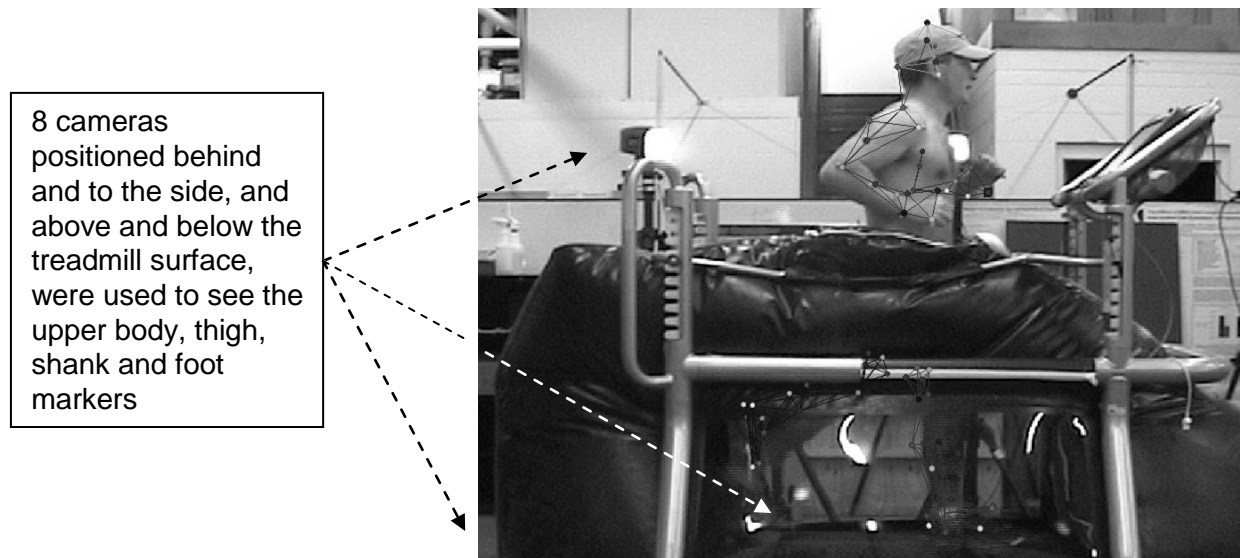
Figure 1: Experimental setup illustrating a subject running in the AlterG treadmill set at 40% of body weight with an overlay of the tracked markers joined to form a stick diagram.

Following a warm-up, participants performed one 3 s static-standing trial while in the anatomical standing position with the elbows flexed at 90° and the forearms pronated so that the palms of the hands faced down. Markers from the medial elbows, knees and ankles, and all from the pelvis region were then removed. Tight-fitting neoprene shorts were then worn, which were zipped into the airtight enclosure around the AlterG treadmill (P200; AlterG, Fremont, CA, USA). Participants performed several trials in this treadmill including one trial at 40% of body weight (unweighted) running at 3.35 m/s (7.5 mph). Participants also performed one running trial at 3.35 m/s on a dual-belt treadmill (weighted) instrumented with three-dimensional strain-gauge force platforms under each belt (TM-09-P; Bertec, Columbus, OH, USA). Participants ran for approximately 2 minutes in each condition with a 30 s trial recorded during this time.

Kinetic data from the force platforms in the Bertec treadmill were amplified (16-bit AM6511; Bertec) and connected to the computer via an A-D board (16-bit NI-USB-6229; National Instruments, Austin, TX, USA). Using Cortex software (v2.0; Motion Analysis Corporation, Santa Rosa, CA, USA) the kinetic data were recorded at 1000 Hz and the three-dimensional motions of the markers were recorded at 200 Hz via eight cameras (4xEagle and 4xEagle-4; Motion Analysis Corporation).

Data were analyzed using custom-written code in Matlab (v2010; Mathworks, Natick, MA). Any small data gaps were linearly-joined, and kinematic and kinetic data were then smoothed using a fourth-order low-pass Butterworth filter at 6 Hz and 18 Hz cut-off frequencies, respectively. Cut-offs were determined from a residual analysis (Winter, 2005). Heel-strike and toe-off events were determined using the force platform data on the normal treadmill rising above and falling below 30N, respectively. These events on the AlterG treadmill were determined as the maximum forward and backward positions of the toe marker. Data from heel-strike to subsequent heel-strike of the same leg (0% to 100% of stride time) were time

normalized to 101 data points using a cubic spline interpolation. Using the thigh and shank clusters and four foot markers (heel, lateral posterior, first and fifth metatarsal), the three-dimensional joint coordinate systems (JCS; Grood and Suntay, 1983) for the thigh, shank and foot were calculated. Five random strides for each participant were extracted.

Variables of stride frequency, and from the JCS normalized to the anatomical standing position (i.e. 0 is angle during standing), the knee flexion-extension (kneeA) and ankle dorsi-plantar flexion (ankleA; -ve dorsiflexion; +ve plantarflexion) angles were calculated. The variability in kneeA and ankleA were determined by calculating the SD at each of the 101 points across the 5 trials and averaging this for each participant. Data were summarized as the average and SD of the six participants. Differences in the stride frequency and variability measures between running conditions were compared using Wilcoxon matched pairs tests at an alpha level of 0.05.

RESULTS: In unweighted treadmill running there was a statistically significantly slower stride rate (67 ± 10 strides/min) than on the normal treadmill (83 ± 5 strides/min; p = 0.03). The leg kinematics demonstrated several notable differences (Figure 2). During stance at approximately 0 to 40% of the stride there is the typical flexion-extension knee angle action and dorsi-plantar flexion ankle angle actions on the normal treadmill. These actions are less pronounced and each participant has more unique patterns in the unweighted condition. During the swing phase at approximately 41-100% of the stride, the knee and ankle patterns observed in the unweighted condition are more typical of normal treadmill running, but again there is still a more unique pattern between participants.
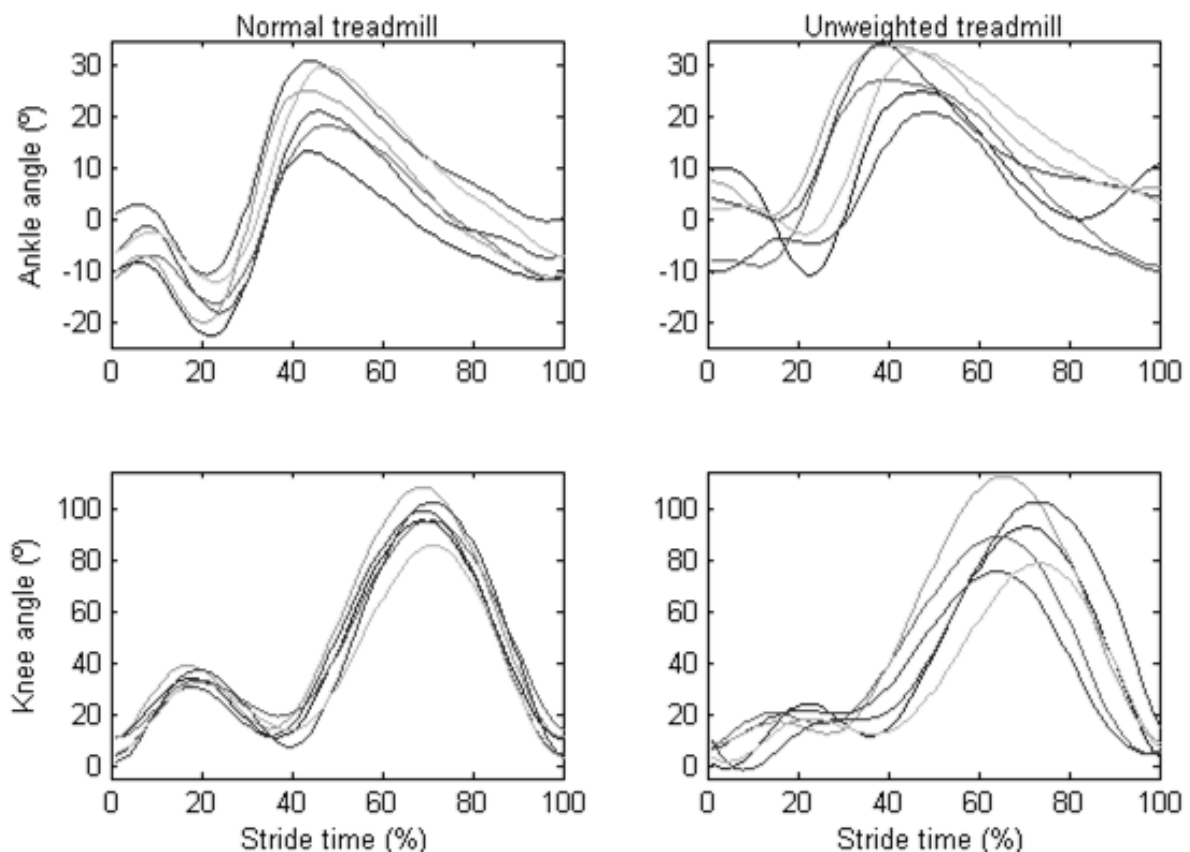

Figure 2: Knee flexion-extension and ankle dorsi-plantar flexion angles for 6 participants running at 3.35 m/s in normal and unweighted to 40% of body weight treadmill conditions. The line for each participant is the average of 5 trials for the right leg from heel strike (0%) to subsequent heel strike (100%). The stance phase is approximately 0-40%.

Variability between participants is quite large, particularly on the unweighted treadmill, as evident in Figure 2. Within-participants variability was smaller, as demonstrated by the average SD for the entire stride (Table 1). For all variables the within-participant variability was significantly greater in the unweighted than in the normal treadmill condition (p<0.05).

Table 1: Variability in leg kinematics for six healthy participants running at 3.35 m/s in normal and unweighted to 40% of body weight treadmill conditions.

	LankleA	LkneeA	RankleA	RkneeA
Normal	0.9 ± 0.2°	1.7 ± 0.2°	0.7 ± 0.1°	1.5 ± 0.3°
Unweighted	1.8 ± 0.6°	3.1 ± 2.1°	1.3 ± 0.6°	2.4 ± 0.7°
p	0.03	0.03	0.03	0.04

Note: L = left; R = right; ankleA = ankle dorsi-plantar angle; kneeA = knee flexion-extension angle. Data are the average ± SD of the SD for 5 trials, across 101 data points for 6 participants.

DISCUSSION: Using a treadmill that simulates a reduction in gravity enables a person to run with a modified gait. The principal differences are that during unweighted treadmill running there is less movement and that there is greater between- and within-participant variability than in normal treadmill running. For astronauts, this information may contribute to equipment design, or use of unweighted treadmill running may improve specificity of training prior to space travel, or assist in rehabilitation after return to earth from prolonged space flight. The less movement may have beneficial implications for the rehabilitative settings where patients would have to work less hard, and, as this may be less painful they may subsequently be able to perform therapeutic exercises that otherwise would not be possible. In addition, the greater variability may be beneficial as the patient is able to use greater combinations of the degrees of freedom to complete the task, and as such is able to avoid the more painful movement patterns. In contrast, it is possible that both the different movement and greater variability leads to an unusual motion that is uncoordinated or exceeds previous limits of motion. Injured runners have been found to decrease variability (Hamill et al., 2000) thus it is unknown if subjecting patients to treatments inducing greater variability through unweighted treadmill running is advised. There is likely a balance between performing more exercise versus performing coordinated exercise that better represent the intended movement pattern. Further research incorporating multi-disciplinary perspectives, including biomechanics, muscle physiology and motor control, and using patients as opposed to healthy participants is necessary to evaluate the best approach to assessing the rehabilitative benefits of devices that induce novel movement patterns.

CONCLUSION: Using a treadmill that unweights a person enables them to run with a modified gait. In comparison to normal treadmill running, during unweighted treadmill running there is less movement and there is greater between- and within-participant variability. The benefits of these changes to training and rehabilitation settings require further research using participants from the targeted population.

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