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How do low horizontal forces produce disproportionately high torques in human locomotion?

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Abstract

Although horizontal ground forces are only ~15% of vertical forces, they account for 47% and 33% of the metabolic cost in walking and running. To explain these disproportionately high metabolic costs we hypothesized that low horizontal ground forces generate relatively high torques on body segments during locomotion and this is mediated by long moment arms. We compared external force moment arms and discreet torques applied to the body segments by horizontal and vertical forces during walking and running. Sixteen subjects (21.9 ± 1.9 years) walked at 1.5 m/s and 10 subjects (23.2 ± 2.0 years) ran at 3.83 m/s. Segmental torques in the sagittal plane were partitioned into components due to horizontal and vertical forces and quantified by their angular impulses. The mean (±S.E.) ratios of horizontal to vertical ground forces (GF ratio) and angular impulses (AI ratio) in walking were 0.131 (±0.003, 95% C.I. 0.124 to 0.137) and 0.530 (±0.018, C.I. 0.497 to 0.569). Results were similar in running. In both gaits the AI ratios were significantly greater than the GF ratios because the respective C.I.s did not overlap. The horizontal forces produced 53% and 41% as much angular impulse on the body segments as did the vertical forces in walking and running despite being only 13% as large. In the two movements the moment arms for the horizontal forces averaged across foot, leg, thigh, and trunk body segments were 3.8 fold larger than those for the vertical forces. The data supported the hypothesis and suggest that the relatively low horizontal vs vertical forces accounted for a disproportionately higher percentage of the angular impulses placed on the body segments and this effect was due to relatively long moment arms for horizontal forces. These results partially explain the relatively large metabolic cost of generating relatively low horizontal forces.

Keywords

walking and running biomechanics; ground force; inverse dynamics; metabolic cost

INTRODUCTION

In human locomotion vertical forces are often five fold larger than horizontal forces (Bates et al., 1983; Bresler et al., 1950; Cavanagh et al., 1980; Kram et al., 1998; Munro et al., 1987). These external forces create torques around each segment's mass center via their moment arms (Bresler et al., 1950; Elftman, 1939; Elftman, 1940; Winter, 1980; Winter, 1983a). Moment

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arms for external forces are often much longer than those for muscle forces (Biewener et al., 2004). Therefore, relative to ground and joint forces, muscles produce high forces to create torques and these muscle forces generate a significant metabolic demand (Cavagna et al., 2000; Chang et al., 1999; Farley et al., 1992; Gottschall et al., 2003; Griffin et al., 2003; Roberts et al., 1998). Curiously, the relatively low horizontal forces account for a disproportionately high amount of this metabolic demand. Through a series of experiments in which aiding and impeding horizontal forces were applied, Gottschall and Kram (Gottschall et al., 2003) showed that horizontal ground forces in walking were associated with 47% of the metabolic cost and Chang and Kram (Chang et al., 1999) showed that horizontal ground forces in running were associated with 33% of the metabolic cost.

Humans walk and run with the joints of the lower extremity near the anatomical position throughout the stance phase and most body segments approximate the vertical direction (Derrick et al., 1998; Grasso et al., 2000; Kuster et al., 1995; Leroux et al., 2002; Redfern et al., 1997). Because of this orientation, it is likely the moment arms for horizontal forces are longer than moment arms for vertical forces. The longer and shorter moment arms for horizontal and vertical forces should respectively increase and attenuate the mechanical and metabolic cost of these forces. The relative moment arm lengths for horizontal and vertical forces may therefore provide the underlying mechanism for the relatively high metabolic cost of generating relatively low horizontal ground forces. Namely, despite their low magnitude, horizontal forces may produce high external torques on each body segment.

We now hypothesize low horizontal ground forces generate relatively high mechanical loads (i.e. torques) on our body segments during locomotion and this effect is mediated by long moment arms for the horizontal forces. The purpose of this study was to compare external force moment arms and the discreet torques applied to body segments by horizontal and vertical forces during level walking and running. With this work we attempt to integrate the physiological and biomechanical demands of human locomotion by investigating the theoretical relationship between mechanical loads and previously reported metabolic costs.

METHODS

Participants

After providing written informed consent in accordance with University procedures, two independent samples of 16 and 10 subjects were tested in level walking and level running conditions, respectively (Table 1). All subjects were apparently healthy with no previous lower extremity injuries, musculoskeletal diseases, or neurological impairments.

Measures and Instrumentation

Horizontal (i.e. anteroposterior) and vertical ground forces for walking and running were obtained at 1000 Hz and 960 Hz respectively using a force platform (LG6-4-1, AMTI, Newton, MA) imbedded in a 20 meter walkway. Sagittal plane kinematics were collected during walking with a Panasonic VHS videocassette recorder at 60 Hz (model AG-450) and during running with a Qualisys ProReflex MCU 240 camera at 120 Hz. An infrared timing system (model 63520, Lafeyette) was used to measure the time required for each subject to complete the trial.

Procedures

Testing for each subject was conducted in one session lasting approximately 75 minutes. Subjects wore a T-shirt, spandex bicycle shorts, and athletic shoes. Six reflective markers designating the ends of selected body segments were placed on the right side of the body: 1) on the lateral side of the 5th metatarsal head, 2) in line with the metatarsal head at the heel of the shoe, 3) the lateral malleolus, 4) the knee joint center of rotation, 5) the greater trochanter, and 6) on the lateral neck. Subject height, weight and lower extremity segmental circumferences were recorded. Walking and running speeds were constrained to 1.5 m/s + 5% and 3.83 m/s + 5%, respectively. A gait trial was retained if the speed was within the target range, the right foot was completely on the force platform and the subject did not alter his or her gait to strike the force platform. A total of 5 successful trials were collected per subject and condition.

Data Analysis

Mechanical loads on body segments were determined through force platform measurements and inverse dynamic modeling. Inverse dynamics predicts inertial torques, gravitational torques, joint forces, and joint torques on each segment. For the purposes of our study, only the loads due to measured ground and predicted joint forces were investigated because these represent the external mechanical loads applied to body segments in horizontal and vertical directions. Inertial, gravitational, and joint torques calculated through inverse dynamics were not included in this analysis. The inertial torques produced less than 1% of the total external angular impulse on each segment and were also not directly attributable to distinct horizontal and vertical forces. The gravitational torque was eliminated from this analysis because we calculated the torques around the segmental mass centers as opposed to the joint centers. This omission did not however change the ratio of angular impulses from the horizontal and vertical forces described below. Had we calculated torques at each joint, the horizontal force moment arms would increase more than the vertical force moment arms due to the more vertically oriented segment positions and this would increase the angular impulses from the horizontal forces. However, the angular impulses from the gravitational force would increase the total angular impulses from vertical forces to keep the ratios as they are reported here. Joint torques quantify the muscular responses to the external torques produced by the applied horizontal and vertical forces during periods of lengthening contractions and they produce these external torques and forces during periods of shortening contractions. For our purposes, the net joint torques were not considered as external loads. They were in fact nearly equal in magnitude and opposite in direction to the torques from the ground and joint forces.

Cartesian Coordinates of the reflective markers were digitized from kinematic records using Peak 5 Motion Analysis (Peak Performance Technologies, Englewood, CO) for walking and QTrac software (Innovision Systems Inc., Columbiaville, MI) for running. High-frequency error from the digitizing process was removed using a low-pass Butterworth digital filter with an optimization routine to select the particular cut-off frequency (Winter, 1990) which averaged 5.0 Hz for all subjects and joint markers. The lower extremity and trunk were modeled as linked system of rigid segments and inverse dynamics were used to calculate vertical and horizontal joint reaction forces on the foot, leg, thigh, and trunk segments (i.e. the joint reaction forces at the ankle, knee, and hip). The magnitude of the segmental masses, their moments of inertia, and the locations of the mass centers were estimated from the position data using a mathematical model (Hanavan, 1964), relative segmental masse values (Dempster, 1959), and individual anthropometric data. Most relevant to the present study, these techniques predicted segmental mass centers to be ~43% of the segment lengths from the proximal segment ends. The ground forces acting on the foot and the joint forces on all segments were used to assess the external, horizontal and vertical mechanical loads on each segment by calculating the torques produced on each segment by these forces in each kinematic frame throughout the stance phase. Moment arms were calculated for each force as the perpendicular distance between the force vectors and the segmental mass centers. In practice, the moment arms were determined by calculating the horizontal and vertical distances between the segment endpoints and the segment center of masses. Angular impulses were calculated as the area under each torque-time curve and used to assess the total angular load by each force on each body segment. The sums of the angular impulses due to horizontal and vertical forces were determined for each segment and for the entire body. The angular orientations of the foot, leg, thigh, and trunk body segments were also calculated throughout stance. The angular orientation for each segment and the length of each moment arm was averaged over the stance phase to provide appropriate descriptors of these variables. These mean values were then averaged across trials and subjects and they are presented along with their standard errors.

Statistical Analysis

We tested the hypothesis by comparing the ratio of angular impulses generated by the horizontal and vertical forces (AI ratio) to the ratio of the mean horizontal and vertical ground forces (GF ratio) during the stance phase. We calculated these ratios in all trials for each subject and then calculated subject means from the trial values. The mean ratios were calculated across all subjects along with the 95% confidence intervals (C.I.). The confidence intervals were compared against each other to determine whether the C.I. for the applied torques was significantly larger ($p<0.05$) than the C.I. for the ground forces. Walking and running gaits were analyzed independently.

RESULTS

Walking

The GF ratio was 0.131 with a standard error (S.E.) of 0.003 (Figure 1a and Figure 2). The 95% CI for this ratio was 0.124 to 0. 137. The mean (± S.E.) angular orientations from the horizontal were 12 ± 1°, 75 ± 1°, 93 ± 1°, and 86 ± 1° for the foot, leg, thigh, and trunk segments throughout the stance phase, respectively. These data indicated the foot was more horizontally oriented while the other segments were more vertically oriented. The mean moment arm lengths at the foot were similar for vertical and horizontal forces but were dissimilar for the other segments (Figure 3a). The mean moment arms for horizontal forces were 2.9, 4.8, and 12.9 fold longer than the corresponding vertical force moment arms for the leg, thigh, and trunk segments, respectively. The horizontal force moment arms tended to increase from distal foot to proximal trunk while those for the vertical forces decreased.

The horizontal force angular impulse was lowest at the foot and increased continuously through the more proximal body segments with the angular impulse being 3.0 fold larger at the trunk than at the foot (Figure 4a). The vertical force angular impulses showed the opposite trend and had decreasing values from foot to trunk with the trunk impulse being 0.3 fold as large as the foot impulse. The total mean (± S.E.) horizontal and vertical angular impulses were 30.8 ± 2.1 and 59.1 ± 4.9 Nms. Respectively, horizontal and vertical forces produced 34% and 66% of the total angular impulse. The observed mean (± S.E.) AI ratio was 0.53 (± 0.018) and the 95% confidence interval for this mean ratio was 0.497 to 0.569. The 95% C.I.s for AI ratio and GF ratio did not overlap and therefore indicated the AI ratio was significantly larger than the GF ratio. In fact, the AI ratio was 4.1 times as large as the GF ratio.

Running

The GF ratio was 0.138 with an S.E. of 0.002 (Figure 1c and Figure 2). The 95% CI for this ratio was 0.134 to 0.142. The mean (± S.E.) angular orientations from the horizontal were 14 ± 2°, 66 ± 1°, 97 ± 2°, and 81 ± 3° for the foot, leg, thigh, and trunk segments throughout the stance phase, respectively. These data indicated the foot was more horizontally oriented while the other segments were more vertically oriented as observed in walking. The mean moment arm lengths at the foot were similar for vertical and horizontal forces but were dissimilar for the other segments (Figure 3b). The mean moment arms for horizontal forces were 2.1, 3.5, and 6.3 fold longer than the corresponding vertical force moment arms for the leg, thigh, and trunk segments, respectively.

The horizontal force angular impulse was lowest at the foot and increased continuously through the higher body segments with the trunk angular impulse being 3.0 fold larger than the foot impulse (Figure 4b). The vertical force angular impulses showed the opposite trend and had decreasing values from foot to thigh. The trunk angular impulse was larger than the thigh angular impulse, but was only 0.6 fold as large as the foot angular impulse. The total mean (± S.E.) horizontal and vertical angular impulses were 20.2 ± 4.5 and 50.3 ± 12.0 Nms. Horizontal forces produced 29% of the total angular impulse and the vertical forces produced 71% of the total angular impulse. The observed mean (± S.E.) AI ratio was 0.405 (± 0.013) and the 95% confidence interval for this mean ratio was 0.381 to 0.430. The 95% C.I.s for the AI ratio and GF ratio did not overlap and therefore indicated the AI ratio was significantly larger than the GF ratio. In fact, the AI ratio was 2.9 times as large as the GF ratio.

Summary

The mean horizontal ground and joint forces averaged 13% of the mean vertical ground and joint forces over both movements. In contrast, the mean moment arm lengths for all horizontal forces averaged 381% of the corresponding values for vertical forces. The interaction of the forces and moment arms enabled the low horizontal forces to produce 47% as much of the total angular impulse as produced by the vertical forces and 32% of the total angular impulse on the body segments.

DISCUSSION

The purpose of this study was to compare external force moment arms and the discreet torques applied to the body segments by horizontal and vertical forces during level walking and level running. We attempted to explain the relatively high metabolic cost of relatively low horizontal forces previously observed in walking (Gottschall et al., 2003) and running (Chang et al., 1999) by showing these low forces create disproportionately high mechanical loads (i.e. angular impulses) on body segments through their long moment arms. It is well established that the metabolic cost of locomotion is due to the generation of muscular force (e.g. (Grabowski et al., 2005; Griffin et al., 2003; Kram et al., 1990; Roberts et al., 1998) and that muscle forces are generated during locomotion in direct proportion to external torques applied to body segments (Alexander, 1991; Biewener et al., 2004; Elftman, 1939; Winter, 1980; Winter, 1983b). Through this logic we therefore show our underlying assumption that external torques applied to our body segments could explain previously reported metabolic cost of locomotion is reasonable. We are certainly not the first investigators to relate lower extremity torques to metabolic costs in locomotion (e.g. (Alexander, 1991; Biewener et al., 2004). The novelty of our work however lies in partitioning external torques into those generated independently by horizontal and vertical forces. This partitioning provided a more precise evaluation of the functional outcome for the ostensibly low horizontal and high vertical forces.

In each movement the leg, thigh, and trunk were oriented closer to the vertical than horizontal direction throughout stance. These orientations created longer moment arms for horizontal forces and shorter moment arms for vertical forces. They also reduced the overall mechanical cost of locomotion by reducing external torques applied onto the segments by the large vertical forces and therefore the muscular responses needed to produce the tasks. In normal walking the upright posture directs the resultant ground force relatively close to skeletal joint centers reducing the moment arms of this force to the ankle, knee, and hip joints (Biewener et al., 2004). More directly, our results showed this posture created short moment arms for large vertical ground and joint forces. This outcome also explains why both obese and older adults walk with more erect postures than lean or young individuals (Browning et al., 2005; DeVita et al., 2000a; DeVita et al., 2003). The erect postures reduced the mechanical loads generated by large vertical forces and thus attenuated the effects of excess weight in obese and muscle weakness in old adults. We previously observed that a more erect posture in old compared to young adults enabled them to use the skeletal system more and muscular system less to maintain vertical support (DeVita et al., 2000b). Scaling body posture and the length of the moment arm for the ground force to body mass has also been observed across eight animal species with approximately a 3,000 fold range in mass (Biewener, 1989). The larger animals reduced the external moment arm for the ground force by running with more erect postures which then reduced muscle forces needed to support the animals. Interestingly, the more erect and vertical postures used by larger, heavier animals should increase the metabolic and mechanical costs of the horizontal forces through their longer moment arms in this posture. In running however the body segments, while still being primarily oriented near vertical, more closely resemble the posture seen in bent-hip, bent-knee walking (Adams et al., 1999; Grasso et al., 2000). In running compared to walking, the muscle mechanical advantage at the knee is decreased due to a more flexed knee position which partially explains the greater metabolic cost for running (Biewener et al., 2004) and the lower ratio between horizontal to vertical angular impulses presently observed in running compared to walking. Thus we can partially attribute the greater metabolic cost of running compared to walking to the relatively larger mechanical demand of vertical forces in this gait.

Gottschall and Kram (Gottschall et al., 2003) found horizontal forces are associated with 47% of the total metabolic cost in level walking, indicating horizontal forces had approximately the same metabolic cost as vertical forces despite being much lower in magnitude. We showed horizontal forces accounted for 34% of the total angular impulse placed on the body in level walking. Combining our results with those of Gottschall and Kram suggests that on a per unit force basis, horizontal forces were about four times more expensive in terms of metabolic cost. This result could be partially explained by the passive transmission of vertical forces through bones of the lower extremities, particularly during midstance. Despite large vertical forces generating external torques on the body segments, as much as 50% of the force is passively transmitted through skeletal bones (Anderson et al., 2003). However, during the early and late portions of stance, less force is directed longitudinally through the bones. It is during this time that large amounts of muscle activity are required to provide support against the vertical forces (Sasaki et al., 2006) and to resist the torques these forces produce.

Chang and Kram (Chang et al., 1999) found that the metabolic cost associated with horizontal forces in level running was about 33% of the total cost. They deduced that per unit of force generated, the horizontal propulsive forces were 3.8 times more expensive in metabolic cost than vertical forces. We showed horizontal forces accounted for 29% of the total angular impulse placed on the body in running which is seemingly quite similar to Chang and Kram’s metabolic ratio. However, the horizontal forces only generated 2.1 times more angular impulse than the vertical forces per unit of force. Therefore the vertical forces contribute more to the angular impulse than to the metabolic cost in level running, as was the case in level walking. The close agreement between Chang and Kram’s metabolic cost and our mechanical load supports our concept that the relatively high metabolic cost of generating horizontal forces in running is due to the total external torque and summed angular impulse created by horizontal forces. Overall, we showed horizontal forces accounted for a larger proportion of the total angular impulse in walking (34%) compared to running (29%). The larger value for walking agrees with the studies by Kram and colleagues that showed the metabolic cost of generating horizontal forces was higher in walking (47%) compared to running (33%). The larger relative mechanical load for the horizontal forces in walking compared to running was due to the more vertically oriented body segments and longer moment arms for horizontal forces in walking.

CONCLUSIONS

The near-vertical orientation of the leg, thigh, and trunk created long moment arms for horizontal forces and short moment arms for vertical forces applied to these segments. Across the two gaits the moment arms for horizontal forces were 3.8 fold larger than those for vertical forces. This enabled the horizontal forces to produce, 53% and 41% as much angular impulse on the body segments as did the vertical forces in walking and running despite being only 13% as large. The data supported the hypothesis that the low horizontal ground forces generate relatively high mechanical loads (i.e. torques) on our body segments during locomotion and this effect is mediated by the long moment arms for the horizontal forces. In conclusion, the results partially explain the relatively large metabolic demand of generating relatively low horizontal forces. There do appear to be however other factors involved with the issue of high metabolic costs for low horizontal forces, particularly in walking gaits.

Acknowledgements

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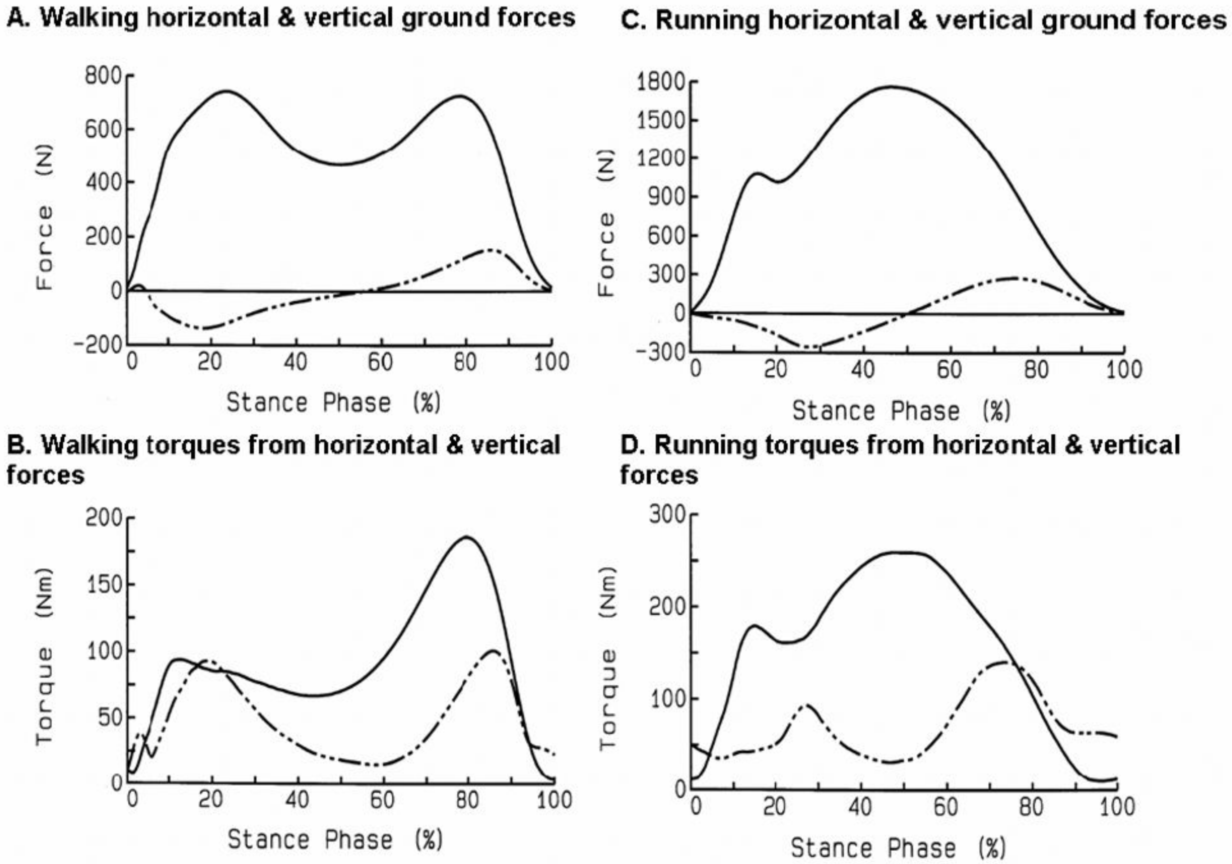

Figure 1.

Top row: horizontal (dashed) and vertical (solid) mean ground force curves averaged across all subjects and trials in walking (a) and running (c). Bottom row: summed total torque curves due to the horizontal (dashed) and vertical (solid) forces averaged across all subjects and trials in walking (b) and running (d).

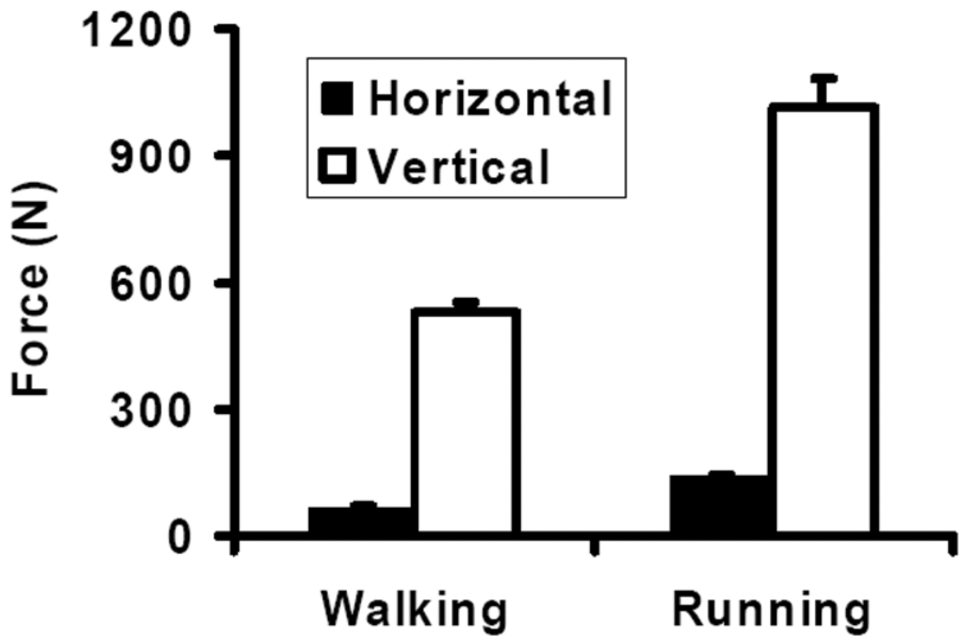

Figure 2.
Mean (+ S.E.) horizontal and vertical ground forces through the stance phases of walking and running.

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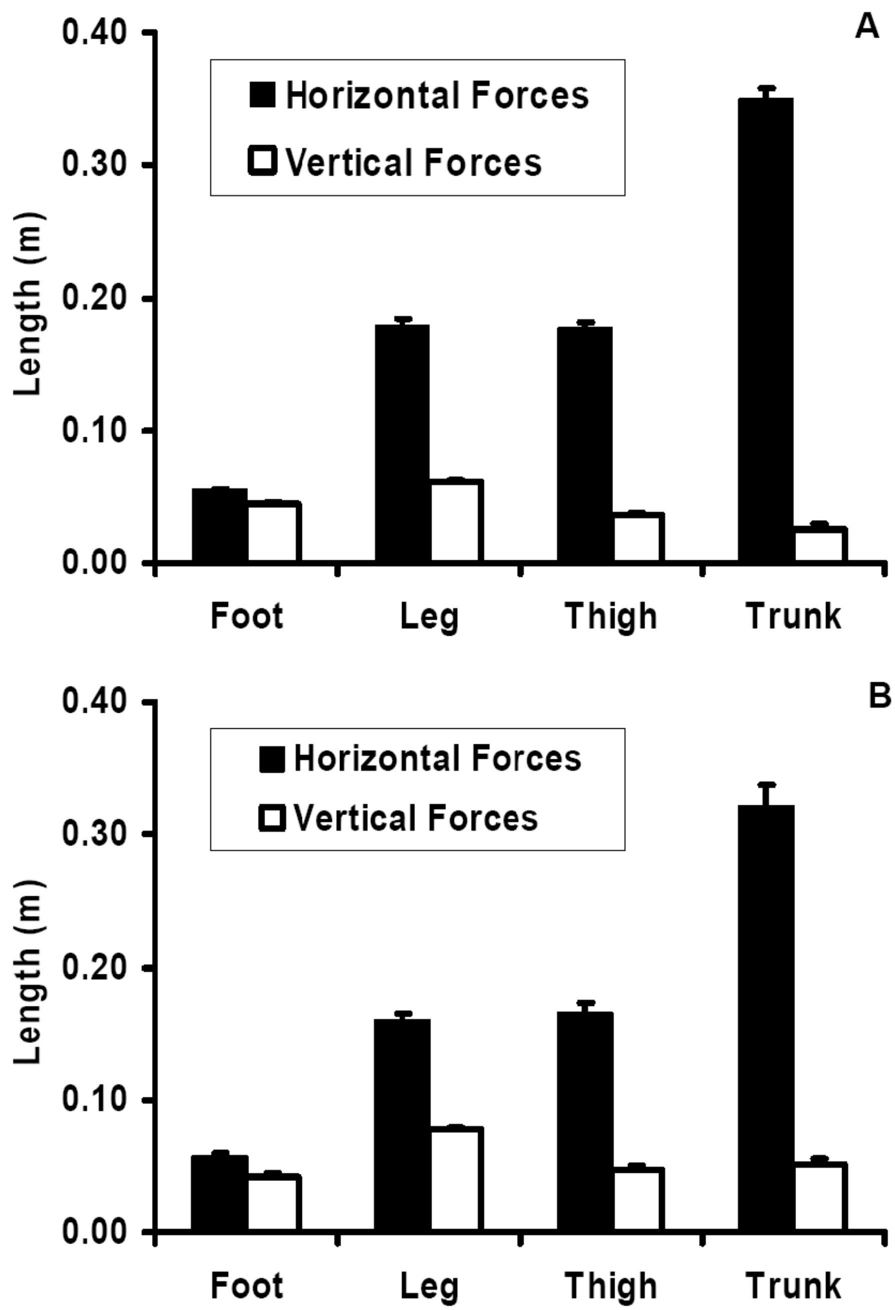

Figure 3.
Mean (+ S.E.) moment arm lengths for the horizontal and vertical forces in walking (A) and running (B) at the foot, leg, thigh, and trunk segments.

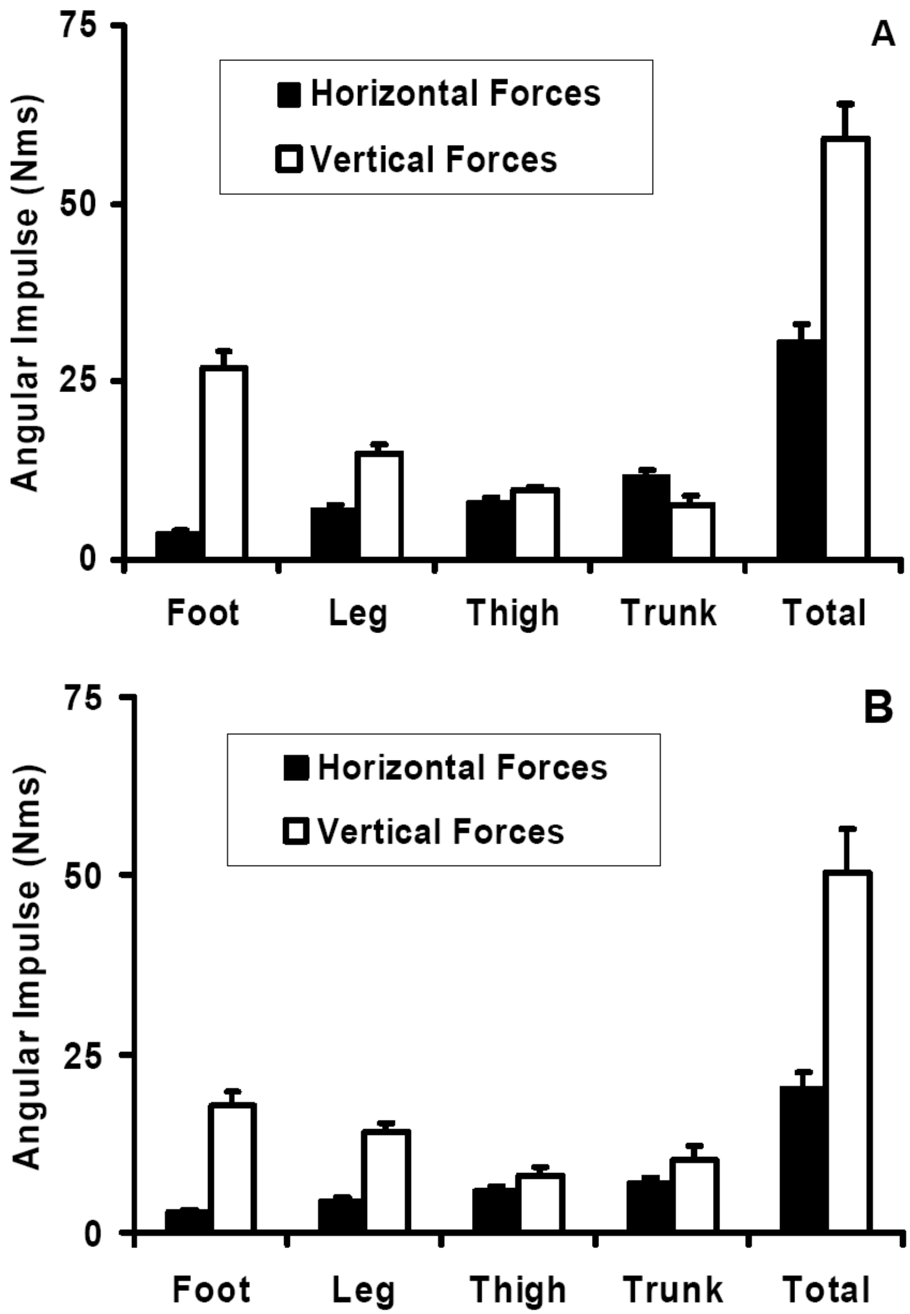

Figure 4.
Mean (+ S.E.) absolute angular impulses due to horizontal and vertical forces in walking (A) and running (B) at the foot, leg, thigh, and trunk segments and their total sum.

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Table 1

Physical Characteristics of the subjects (mean ± sd)

Group	n	Age (yrs.)	Height (m)	Mass (kg)	BMI (kg/m^2)
Walking	16	21.9 ± 1.9	1.70 ± 0.10	65.2 ± 12.2	22.4 ± 3.0
Running	10	23.2 ± 2.0	1.71 ± 0.11	67.4 ± 15.8	22.9 ± 3.1