The Effect of Manipulating Subject Mass on Lower Extremity Torque Patterns During Locomotion

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INTRODUCTION

During locomotion, humans adapt their motor patterns to maintain coordination despite changing conditions (Reisman et al., 2005). Bernstein (1967) proposed that in addition to the present state of a given joint, other factors, including limb inertia and velocity, must be taken into account to allow proper motion to occur.

During locomotion with added mass counterbalanced using vertical suspension to maintain body weight, vertical ground reaction forces (GRF's) increase during walking but decrease during running, suggesting that adaptation may be velocity-specific (De Witt et al., 2006). It is not known, however, how lower extremity joint torques adapt to changes in inertial forces.

The purpose of this investigation was to examine the effects of increasing body mass while maintaining body weight upon lower-limb joint torque during walking and running. We hypothesized that adaptations in joint torque patterns would occur with the addition of body mass.

METHODS

Vertical GRF was measured while ten subjects (5M/5F) walked (1.34 $m \cdot s^{-1}$) and ran (3.13 $m \cdot s^{-1}$) on a Kistler Gaitway treadmill (Amherst, NY). Sagittal plane kinematics were obtained using an optical motion capture system (Smart Elite System, BTS Bioengineering Spa, Milanese, IT).

Subjects completed trials with 5 added mass (AM) conditions (0%, 10%, 20%, 30% and 40% of body mass) applied in random order. The added mass was achieved by having subjects wear a weighted vest (X-Vest, Perform Better, Cranston, RI). Body weight was maintained using a pneumatic unweighting system (H/P/Cosmos Airwalk, Nussdorf-Traunstein, Germany).

Ten consecutive strides were analyzed after each subject achieved steady-state within each trial. Left hip and knee joint torques were computed using inverse dynamics. Positive torques represent hip and knee extension. Positive and negative angular impulse (AI) for each joint was found during the stance and swing phase of each stride, respectively. Trial means for each variable were computed for each condition.

Analysis of variance with repeated measures was used to determine the affect of AM conditions on angular impulse. Separate analyses were conducted for walking and running. Tukey-Kramer post-hoc tests were used to determine differences when a significant AM effect ($p<0.05$) was found.

RESULTS AND DISCUSSION

For all conditions, knee flexor torque dominated during the early stance phase of walking (see Figure 1), which is contrary to the presence of extensor torque reported in literature (De Vita et al., 1996). This may be a specific characteristic of treadmill locomotion that is not apparent during overground gait.

During the stance phase, hip extensor AI increased with AM during walking, but was not affected during running. Hip flexor AI increased with AM during running, but was unaffected during walking. Knee extensor AI was not affected by AM, but knee flexor AI increased with AM during running.

During the swing phase, AM did not affect hip AI during walking, but both flexor and extensor AI decreased with increasing AM during running. Knee extensor AI decreased with increasing AM during both walking and running. Knee flexor AI during swing was unaffected by AM in either gait mode.

SUMMARY/CONCLUSIONS

Our findings suggest that when mass is increased while maintaining body weight, adaptations in joint torques occur that are specific to each locomotion style (walking or running). Hip and knee AI were affected by AM, but the effects were different between walking and running, and were inconsistent. It appears that the control system adapts differently to walking with AM than running.

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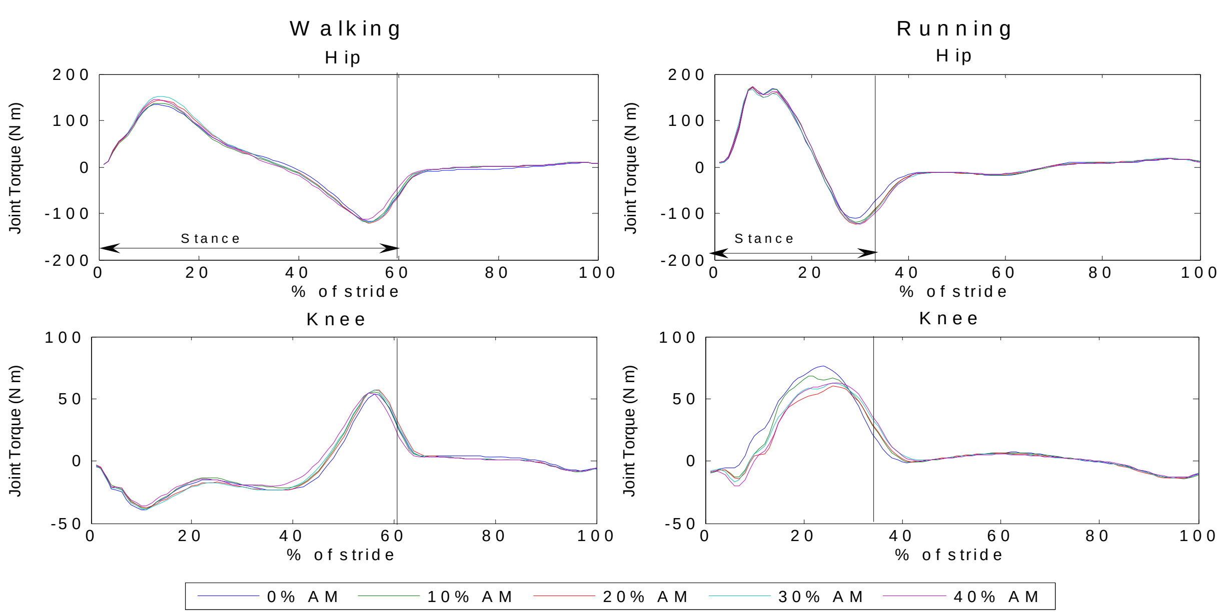

Figure 1. Hip and knee joint torque trajectories during walking and running. Extensor torques are positive and flexor torques are negative.