AMPUTEES USE LESS JOINT TORQUE COVARIATION THAN ABLE-BODIED SUBJECTS TO GENERATE LEG FORCE DURING WALKING

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INTRODUCTION

An excess of degrees-of-freedom (DOF) within the locomotor system, which we term motor abundance, allows human walkers to select from many equally successful joint coordination strategies to achieve consistent gait. Implicit gait goals are made consistent (ie. stable) over many repetitions through selection of goal-equivalent joint-torque combinations [1-4]. Walkers can achieve this stability through multiple strategies, including (1) covariation (CoV), in which ankle, knee, and hip joint torques coordinate to compensate for joint torque deviations, or (2) individual variation (InV), in which one or more joints directly influence leg force such that consistent (or variable) joint torques lead to stable (or modulated) leg force over many steps. We were able to isolate the contributions of these two possible stabilization strategies by applying a modified uncontrolled manifold (UCM) analysis [5]. Previous work has shown that healthy humans change CoV and InV contributions as a hopping task became more difficult [5].

Locomotor control may also change after neuromuscular injury or impairment makes walking more difficult [6]. Traumatic transtibial amputation results in the isolated removal of active ankle control, thus effectively removing a degree of freedom on the affected side. Due to this reduction, we hypothesized amputees would demonstrate a limited ability to covary joint torques in their amputated leg, instead relying on individual variation of the remaining intact knee and hip torques to generate leg force.

METHODS

Six male subjects gave written informed consent before walking on an instrumented dual-belt treadmill for two minutes at 75% of preferred walking speed (PWS, 1.22±0.098 m/s). Three subjects had a traumatic transtibial right leg amputation over one year prior to our data collection. All amputee subjects walked using their own custom made and appropriately fit prosthetic legs. Three able-bodied controls were gender, body weight (83.4±14.1kg), and leg length (92.3±5.3cm) matched to each amputee subject. Ground reaction forces were collected independently for each limb (1080Hz, AMTI, Fig 1), while simultaneous kinematics data were captured using a six-camera motion analysis system (120 Hz, VICON). Data from the right leg of control subjects (CON), sound side of amputees (Sd), and prosthetic side of amputees (Px) were analyzed separately (92±6 steps per leg).

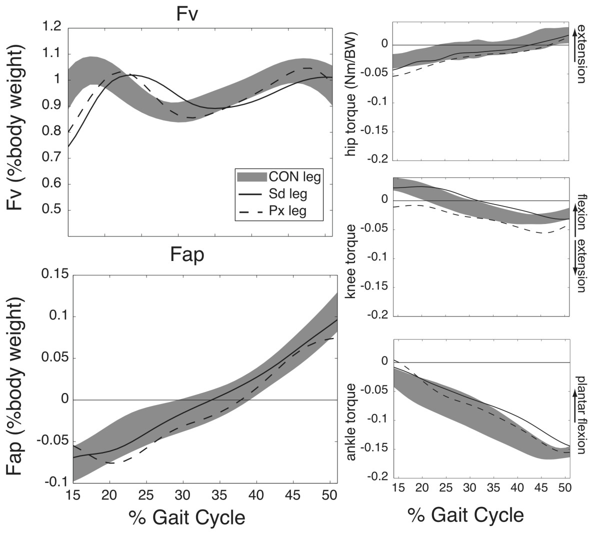

Figure 1: Mean F_v and F_{ap} and joint torques for each leg type over single leg stance at 75% PWS. Gray regions indicate ±2 standard deviations of the CON leg data.

A 3-DOF uncontrolled manifold (UCM) analysis was applied to quantify the structure of inter-step joint-torque variance to generate consistent (stable) leg force values [7]. Inter-step variance of joint-torques was partitioned into orthogonal components that were goal-equivalent for leg force (GEV) and non-goal-equivalent for leg force (NGEV). We used the normalized difference between these two variance components, called the index of motor abundance (IMA), to characterize the use of motor abundance to stabilize net force [4]. Separate analyses were conducted and two IMAs calculated for stabilization of vertical (F_v) and anterior-posterior (F_{ap}) leg forces at every 1% of stance phase. Mean IMA values throughout single-leg stance across all subjects were evaluated for significant differences from zero using a two-tailed Student's t-test (α=0.05). An IMA greater than zero indicates the ankle, knee, and hip torques were coordinated to generate the same leg force in each step, which we interpret as purposeful stabilization. An IMA less than zero indicates active modulation such that intra-leg joint torques purposefully combined to produce different leg forces with each step. An IMA not different from zero indicates no structure to leg force generation.

To isolate the contribution of two possible stabilization strategies, we performed an additional modified UCM analysis [5]. In this analysis we calculated two additional stabilization metrics, InV and CoV. InV is the IMA value for a surrogate data set in which all joint torque covariation has been removed. CoV is the IMA value that isolates the effects of covariation between the joints and is calculated as the difference between IMA and InV. CoV and InV can be interpreted similarly to IMA, where values greater than zero indicate contribution of that strategy to consistent, stabilized leg force, while values less than zero indicate a contribution to modulate leg force with each step. We tested for significant differences from zero for IMA, InV, and CoV for all leg types (CON, Sd, and Px) using a two-tailed Student's t-test (α=0.05). We hypothesized that Sd legs would have similar IMA, InV, and CoV values to CON legs, while Px legs were expected to show no CoV contribution to leg force stabilization (IMA>0) or modulation (IMA<0).

RESULTS AND DISCUSSION

Able-bodied walkers use covariation to generate leg force during single leg stance. At 75% PWS, CON leg joint torques covaried (CoV>0, p=0.035, Fig 2A) to minimize the effect of ankle torque variance on F_v, resulting in unstructured IMA values (IMA=0, p=0.124). Able-bodied subjects stabilized F_{ap} (IMA>0, p=0.009) primarily through covariation of ankle, knee, and hip joint torques (CoV>0, p=0.025, Fig 2B). F_{ap} stabilization indicates that leg force generation is consistent throughout two minutes of walking.

Amputees demonstrate a limited ability to covary, instead generating leg force through individual joint torque variance control. Amputee subjects demonstrate no significant Sd or Px CoV for F_v generation (CoV=0, p>0.05), consistent with our expectation that amputees would have a limited ability to covary. In both their Sd and Px legs, amputee walkers modulate F_v with each step (IMA<0, p<0.05) through independent joint torque variance (InV<0, p<0.05, Fig 2A). Initial analysis revealed that F_v was most sensitive to ankle torque throughout single-leg stance (Fig 2C), and F_v IMA trajectories mirror the amplitude of ankle variability throughout single leg stance. These results imply that Sd and Px F_v modulation results from ankle torque deviations.

Amputees appear to utilize asymmetric control strategies in their Sd and Px legs for F_{ap} stabilization. The Sd leg mirrors CON legs, stabilizing F_{ap} (IMA>0, p<0.05) primarily through joint torque covariation (CoV>0, p<0.05, Fig 2B). As predicted, Px legs appear to have limited ability to covary, never demonstrating CoV values statistically different from zero (p=0.925). Amputees instead utilize an independent variance control strategy to maintain consistent F_{ap} generation (InV>0, p=0.028, Fig 2B). Px leg F_{ap} stabilization is likely achieved through highly consistent (ie. low variance) torque generation in the remaining intact hip and knee joints, which F_{ap} is highly sensitive to (Fig 2D).

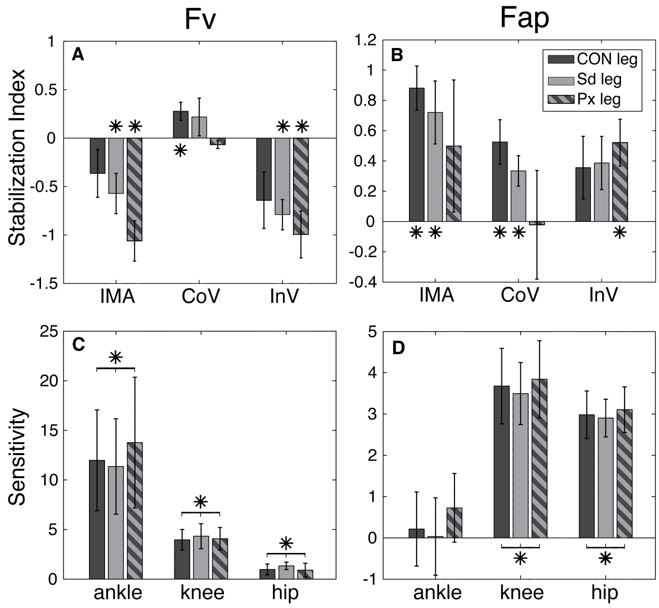

Figure 2: Mean stabilization indices (A and B) and mean sensitivity magnitude (C and D) over single leg stance. Error bars show inter-subject standard deviations. Asterisks indicate values statistically different from zero (p<0.05).

CONCLUSIONS

Individuals with unilateral transitibial amputations used different leg force control strategies compared to able-bodied matched controls during treadmill walking despite reduced biological redundancy. While able-bodied subjects demonstrated inter-joint covariance (CoV strategy), amputees modulated F_v largely through individual ankle torque deviations (InV strategy). Amputees stabilized F_{ap} in their Sd legs using a CoV strategy similar to able-bodied controls, but utilized an InV strategy to generate consistent F_{ap} in their Px legs. Our results show that individuals with transtibial amputations have limited ability to covary joint torques in their affected leg due to reduced redundancy. Amputees must instead rely on controlling individual joint torques to directly influence leg force during single leg stance. The asymmetric leg force control strategies demonstrated by amputee walkers provide further evidence that the human locomotor system may separately adapt and control each leg during walking [8]. A better understanding of how locomotor control strategies change after injury can generalize to improve a range of therapeutic practices.

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