

Control Strategy for Stabilizing Force with Goal-Equivalent Joint Torques is Frequency-Dependent during Human Hopping

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Abstract—Normal human locomotion requires the ability to control a complex, redundant neuromechanical system to repetitively cycle the legs in a stable manner. In a reduced paradigm of locomotion, hopping, we investigated the ability of human subjects to exploit motor redundancy in the legs to coordinate joint torques fluctuations to minimize force fluctuations generated against the ground. Although we saw invariant performance in terms of force stabilization across frequencies, we found that the role of joint torque coordination in stabilizing force was most important at slow hopping frequencies. Notably, the role of this coordinated variation strategy decreased as hopping frequency increased, giving way to an independent joint variation strategy. At high frequencies, the control strategy to stabilize force was more dependent on a direct reduction in ankle torque fluctuations. Through the systematic study of how joint-level variances affect task-level end-point function, we can gain insight into the underlying control strategies in place for automatically counteracting cycle-to-cycle deviations during normal human locomotion.

I. INTRODUCTION

Legged locomotion involves the highly complex and coordinated neuromechanical action of multiple legs, joints, muscles and nerves, the control of which is still not well understood. Diverse groups of running and trotting animals (e.g., Arthropods, Birds and Mammals) follow simple biomechanical rules predicting that these highly complex locomotor systems are controlled such that the body's center of mass mimics a bouncing ball [1-6]. Roboticists have successfully built dynamically stable 1-, 2- and 4-legged running robots based on the simple, task control principles emulated by a spring-mass system [7-8]. Moreover, neuroscientists have found supporting evidence that the central nervous system integrates the enormous amount of sensory information from the periphery into relatively simple neural representations of whole limb function that ascend into the brain [9-11]. Similarly, descending motor commands from the brain appear to contain equally simple, task-relevant commands that are decoded into more complex synergistic muscle activations and movements [12-17]. This mounting evidence suggests the sensorimotor control of complex neuromechanical systems producing complex locomotor behaviors is simplified through a hierarchical reduction in the number of control parameters, which correlate with the relative simplicity of the performance task goals.

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The Uncontrolled Manifold (UCM) concept [18] suggests that motor redundancy in the legs can be exploited to enable achievement of goal-directed locomotor movements. During human locomotion, the biological controller appears to have access to some representation of the biomechanical template for the given locomotor task (e.g., limb function associated with spring-mass dynamics in hopping). For example, leg end-point position and force remains highly consistent over many step cycles despite, and perhaps due to, the seemingly random cycle-by-cycle fluctuations of joint kinematics and torques. We have recently demonstrated in a reduced locomotion paradigm, hopping in place, that joint-level dynamics in human locomotion are coordinated to reduce cycle-to-cycle variance of limb-level function [19-20]. For example, small deviations in joint torques combine in a synergistic manner to minimize deviations in end-point force during hopping [20]. It is not known, however, whether the biological controller accomplishes this by coordinating joint torque deviations to counteract one another, or whether it directly minimizes joint torque deviations. Furthermore, it is unknown how robustly these control strategies are used across different tasks.

The purpose of this study was to determine the extent to which joint torque covariance plays a role in stabilizing forces and to see how this role changes when the task is constrained through a change in hopping frequency. Hopping at increased frequencies reduces the amount of time available to make cycle-by-cycle corrections and involves more extended leg postures, yet the effect of these factors on limb control are unknown. Since it is likely that covariation of the joints would require some level of involvement of heterogenic feedback [19-20], we hypothesized subjects would use coordinated variation of joint torques more at slower frequencies and less at higher frequencies. Understanding the strategies for how humans exploit motor redundancy under different locomotor conditions can have important implications for developing wearable robotic systems that appropriately match natural human behavior.

II. METHODS

A. Subjects

Five male and five female subjects (26.1±4.0 yrs old, 63.3±8.2kg body mass and 170.8±9.6cm height) gave informed consent before participating in this study approved by the Georgia Tech Institutional Review Board. All subjects were healthy and recreationally fit with no history of major musculoskeletal or neuromuscular injury.

B. Equipment

We recorded 3-D lower body kinematics with a 6-camera motion analysis system (120 Hz, Vicon Motion Systems, Oxford, UK) and ground reaction forces and moments with a force platform (1080 Hz, AMTI, Watertown, MA).

C. Experimental Protocol

Subjects hopped in place on their right legs with arms folded across their chests. They matched audible beat of a metronome (see [19-20]). For each trial, the metronome was set to one of three frequencies: 2.2, 2.8, or 3.2 Hz. 2.2 Hz approximates the preferred human hopping frequency [21-22]. Approximately 150-200 hops were collected for each subject at each frequency condition.

III. DATA ANALYSIS

A. Inverse Dynamics

We performed inverse dynamics to calculate sagittal plane torques about the ankle, knee, and hip joints using kinematics and force data (zero-phase 4th-order Butterworth low-pass filter with a 10 Hz cutoff frequency). Inertial properties of the foot, shank, and thigh were estimated based on subject anthropomorphic measurements [23]. For analysis of ground reactions forces alone, we used a 25 Hz cutoff frequency. Data were cropped to include only stance phases, defined by a vertical force above 32 N. We time normalized and expressed the data as a percentage of stance phase. At each 1% of stance phase we calculated the variances of the joint torques and vertical ground reaction force.

B. Uncontrolled Manifold (UCM) Analysis

At each 1% time slice, we studied joint torque variance structure in relation to vertical force (F_v) variance with a derived linear relationship between force and joint torques.

$$[F_v] \approx [S_V(\theta)]T = [s_{va} s_{vk} s_{vh}] \begin{bmatrix} \tau_a \\ \tau_k \\ \tau_h \end{bmatrix} \quad (1)$$

The components of joint torque vector (T) are the sagittal plane ankle (τ_a), knee (τ_k), and hip (τ_h) joint torques. We derived S_V in a previous study (see [19]). Briefly, S_V is a 3x1 matrix and is the transpose of the dynamically consistent generalized inverse of a kinematic Jacobian [24]. The Jacobian relates changes in joint angles ($d\theta$) to a change in vertical endpoint position. S_V take into account leg segment inertias, and is a function of leg joint angles (θ) and the center of pressure (CoP). The components of S_V , that is, the lower-case s 's, can be viewed as the sensitivities of vertical force to each joint torque (a - ankle, k - knee, or h - hip).

We performed an Uncontrolled Manifold (UCM) analysis similar to previous UCM studies [18-20, 25]. At each 1% time slice, we found the average joint torque vector T° across hops, and then calculated ΔT , the deviation of T from T° for each hop cycle. For a given leg posture (θ), there are two orthogonal directions, ε_1 and ε_2 , in which T may deviate from T° without changing the force component applied to the ground. ε_1 and ε_2 are unit-sized torque vectors spanning the null space, or the linearly-approximated UCM. In other

words, ε_1 and ε_2 form a set of basis vectors that represents torque deviations that do not result in a deviation of the force component. We then solved general equation (2). The component of deviation torque ΔT that did not result in a deviation of the force component was shown by (3). The remaining component of ΔT that resulted in a deviation of the force component was shown by (4).

$$0 = S(\theta) \cdot \varepsilon_k \quad (2)$$

$$\Delta T_{//} = \sum_{k=1}^2 \varepsilon_k^T \cdot \Delta T \cdot \varepsilon_k \quad (3)$$

$$\Delta T_{\perp} = \Delta T - \Delta T_{//} \quad (4)$$

The amount of torque variance per degree of freedom that did not contribute to variance of the force component was defined as goal-equivalent variance (GEV, 5). The amount of torque variance per degree of freedom that contributed to variance of the force component was defined as non-goal-equivalent variance (NGEV, 6).

$$GEV = \frac{1}{N} \sum_{n=1}^N \|\Delta T_{//n}\|^2 \cdot \frac{1}{2} \quad (5)$$

$$NGEV = \frac{1}{N} \sum_{n=1}^N \|\Delta T_{\perp n}\|^2 \quad (6)$$

We calculated total torque variance per degree of freedom as (7), where N is the number of hops recorded from a subject hopping at a given frequency condition.

$$TOTV = \frac{1}{N} \sum_{n=1}^N \|\Delta T\|^2 \cdot \frac{1}{3} = (2 \cdot GEV + NGEV) \cdot \frac{1}{3} \quad (7)$$

C. Index of Motor Abundance

GEV and NGEV are not directly comparable across subjects. Therefore, we used the Index of Motor Abundance (IMA, [19-20]) to quantify the degree of force stabilization with one metric. A positive IMA value indicates a joint torque variance structure that stabilizes force, and a negative IMA indicates a variance structure that destabilizes force.

$$IMA = \frac{GEV - NGEV}{TOTV} \quad (8)$$

Isolating the Role of Joint Torque Covariance

It is possible that a joint torque variance structure can arise from unequal variances among the three joint torques, i.e. independent variation, and not necessarily from hop-to-hop deviations of joint torques that cancel out one another, i.e. coordinated variation. To isolate the effects of independent variation, we first found InV, which is the IMA value of a surrogate data set with all possible coordinated variation removed. A surrogate data set was formed from joint torque permutation among all hops of a subject and hopping frequency condition. For example, if 150 hops were collected from a subject hopping at a given frequency, then the surrogate data set consisted of 150^3 hops. Every ankle torque among all hops was combined with every knee and hip joint torque. The surrogate data set had joint torque pairwise covariation values exactly equal to zero, and the variances of the three joint torques were the same as the

original data set. A proof of this general procedure can be found in [26]. Moreover, we verified that each of the joint torque variances was normally distributed using the Lilliefors test for normality [27]. Therefore, UCM analysis on this surrogate data set revealing stabilization or destabilization of force was only sensitive to inequality between variances of the joint torques. If the variances of the joint torques had been equal, then InV would have been 0, which would correspond to no effect of independent variation of joint torques on force.

To calculate the amount of force stabilization or destabilization from coordinated variation alone, we took the difference between the IMA value of the original data set and the InV value as described above to find CoV.

$$\text{CoV} = \text{IMA} - \text{InV} \quad (9)$$

Analogous to the InV metric, the CoV metric considers only the effects of the covariances of the joint torques on the variance of the force component.

IV. RESULTS

A. Variance of Joint Torques and Force

All subjects hopped within 5% of the prescribed frequency. For all conditions, the ankle, knee, and hip joint torques were extensor torques for the majority of stance phase (Fig 1). The resulting ground reaction force had a vertical component trajectory that was uni-modal with a maximum near mid-stance and a horizontal component trajectory that was relatively small and did not follow any particular trajectory (Fig 1).

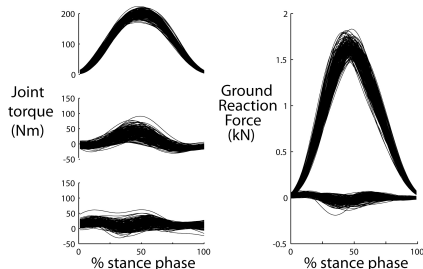


Fig. 1. Sagittal plane hip, knee and ankle joint torques (left, starting from top panel) and ground reaction forces (right) for 175 hops from a representative subject at 2.8 Hz.

Time trajectories of force variances were consistent across frequencies and was bimodal with a local minimum at mid-stance (Fig 2). In contrast, joint torque variances changed qualitatively as subjects hopped more quickly. For 2.2 Hz hopping, trajectories of joint torque variances were unimodal with local maximums near mid-stance (Fig 2). At higher frequencies, joint torque variances decreased, particularly at the ankle. And, like force the ankle torque variance trajectory was bimodal with a minimum near mid-stance.

B. Sensitivities of Force to Individual Joint Torques

Force variance was most sensitive to ankle torque variance and least sensitive to hip torque variance. The dynamically-consistent model we used to map joint torques to force was parameterized by segment inertias, lengths, and angles. The

parameters resulted in a map (S_f) that weighed the influence of the ankle joint on force the most among the three joints. Force was 60% less sensitive to knee torque and 90% less sensitive to hip torque compared to ankle torque.

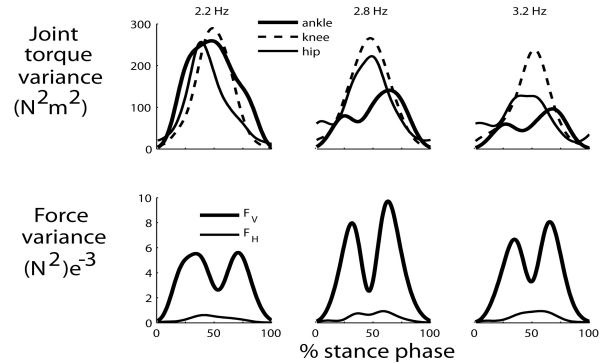


Fig. 2. Representative variances of individual joint torques and forces from same subject as Fig. 1 at each hopping frequency.

C. Utilization of Motor Abundance

Stabilization of force through structuring of joint torque variance was robust across all subjects and frequencies. For all hopping frequencies, the Index of Motor Abundance (IMA) was significantly positive during the beginning, middle, and end of stance phase ($P < 0.005$, Fig 3). The three periods of peak IMA values corresponding to the beginning, middle, and end of stance did not change significantly as subjects hopped more quickly ($P > 0.05$).

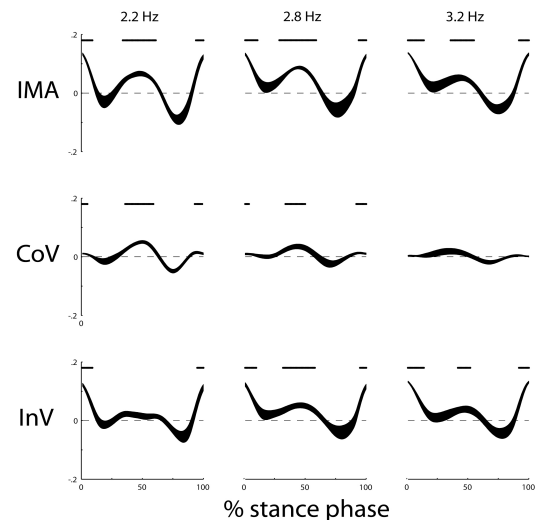


Fig. 3. Index of Motor Abundance (IMA) to stabilize force was consistent across frequencies (top panels) in periods where $\text{IMA} > 0$ (horizontal bars, $p < 0.01$). Contributions to IMA from covariance of joint torques decreased with increasing frequency (CoV, middle panels). Contributions to IMA from the independent variance of joint torques increased with increasing frequency (InV, bottom panels). Data are means \pm 1SE across all subjects.

At 2.2 Hz, both coordinated and independent variation strategies contributed to force stabilization. Coordinated variation stabilized force during mid-stance (Fig 3), while independent variation stabilized force at the beginning and end of stance phase (Fig 3). Coordinated variation explains why force variance reached a minimum at mid-stance despite joint torque variances being at a maximum (Fig 2).

Even though force stabilization was consistent across frequencies (Fig 3), subjects increasingly relied on independent variation of joint torques and less on coordinated variation to stabilize vertical force as they hopped more quickly (Fig 3). Coordinated joint torque variation for force stabilization (CoV) was significantly greater than zero at beginning, middle, and end of stance for 2.2 Hz hopping (0-4, 36-59, 94-100% of stance; $P < 0.005$). At 2.8 Hz, only the beginning and middle of stance phase exhibited significant CoV (0-6, 34-52% of stance; $P < 0.005$). And, at 3.2 Hz we did not find any significant contribution from coordinated variation during stance phase ($P > 0.025$, Fig 3). In contrast, independent joint torque variation for force stabilization (InV) at mid-stance was not significantly greater than zero for 2.2 Hz, but was significant for the higher frequencies ($P < 0.005$; Fig 3).

V. DISCUSSION

Joint torque variance was structured consistently across hopping frequencies to stabilize force. However, the strategy used to structure joint torque variance changed as a function of hopping frequency. At slow frequencies, subjects used interjoint coordination, or coordinated variation, to cancel small hop-to-hop joint torque fluctuations between joints. At high frequencies, hoppers used independent variation, or minimization of ankle torque variance to stabilize force. The UCM analysis with the permutation method as performed here provides an effective tool for quantifying the changing role of interjoint coordination in achieving seemingly invariant motor behavior across changing conditions.

The use of coordinated variation and independent variation to achieve consistent overall performance corresponds to a redundancy of control strategies, in addition to the redundancy of mechanical degrees of freedom, available to the locomotor system. A redundancy of control strategies allows the locomotor system to implement a combination that is most energetically and computationally efficient, and also allows for flexibility in a changing environment. A change in a gait parameter such as hopping frequency yielded a shift in control strategy that did not change vertical force stability. It remains to be seen whether neurological impairments would likewise yield a shift in control strategy that does not compromise performance. This knowledge would yield important insights into effective gait rehabilitation methods and potentially useful robotic control algorithms for wearable prosthetic and orthotic devices. For example, controlling the legs within a goal-equivalent manifold of the task space requires monitoring of relatively few parameters (e.g., end-point position) and allows variability in joint dynamics that may help to promote more effective relearning of gait.

REFERENCES

[1] G.A. Cavagna, N.C. Heglund, and C.R. Taylor, "Mechanical work in terrestrial locomotion: two basic mechanisms in minimizing energy expenditure." *Am. J. Physiol.* vol. 233, pp. R243-R261, 1977
 [2] R. Blickhan, "The spring-mass model for running and hopping." *J. Biomech.* vol. 22, pp. 1217-1227, 1989.

[3] T.A. McMahon, and G.C. Cheng, "The mechanics of running: how does stiffness couple with speed?" *J. Biomech.* vol. 23, pp. 65-78, 1990.
 [4] R. Blickhan, and R. J. Full, "Similarity in multilegged locomotion: bouncing like a monopode." *J. Comp. Physiol. A-Sens. Neural Behav. Physiol.* vol. 173, pp. 509-517, 1993.
 [5] R.J.Full, and D.E. Koditschek, "Templates and anchors: neuromechanical hypotheses of legged locomotion on land." *J. Exp. Biol.* vol. 202, pp.3325-3332, 1999.
 [6] R.J.Full, "Mechanics and energetics of terrestrial locomotion: bipeds to polypeds", in *Energy Transform. Cells and Animals*, W. Wieser and E. Gnaiger, Editors. 1989, Thieme: Stuttgart. p. 175-182.
 [7] M.H. Raibert, H.B. Brown, and S.S. Murthy, "Experiments in balance with a 3D one-legged hopping machine." *Int. J. Robot. Res.* vol.3. pp. 75-92, 1984.
 [8] Ra M.H. Raibert, H.B. Brown, and S.S. Murthy, "3D balance using 2D algorithms?", in *The First International Symposium of Robotics Research*. MIT Press Cambridge. pp. 279-301, 1984.
 [9] R.E. Poppele, , G. Bosco, and A.M. Rankin, "Independent representations of limb axis length and orientation in spinocerebellar response components." *J. Neurophysiol.* vol.87, pp. 409-422, 2002.
 [10] G.Bosco and R.E. Poppele, "Reference frames for spinal proprioception: kinematics based or kinetics based?" *J. Neurophysiol.* vol.83, pp. 2946-2955, 2000.
 [11] G.Bosco, R.E. Poppele, and J. Eian, "Reference frames for spinal proprioception: limb endpoint based or joint-level based?" *J. Neurophysiol.* vol.83, pp. 2931-2945, 2000.
 [12] S. Mori, "Integration of posture and locomotion in adult decerebrate cats and in awake, freely moving cats." *Prog. Neurobiol.* vol.28, pp. 161-195, 1987.
 [13] L.Ting, "Dimensional reduction in sensorimotor systems: a framework for understanding muscle coordination of posture." *Prog. Brain Res.* vol.165, pp. 299-321, 2007.
 [14] L.Ting and J.L. McKay, "Neuromechanics of muscle synergies for posture and movement." *Cur. Op. Neurobiol.* vol.17, pp. 622-628, 2007.
 [15] M.C. Tresch, P. Saltiel, and E. Bizzi, "The construction of movement by the spinal cord." *Nature*, vol.2, pp. 162-167, 1999.
 [16] T. Drew, J. Kalaska, and N. Krouchev, "Muscle synergies during locomotion in the cat: a model for motor cortex control." *J. Physiol.*, vol.586, pp. 1239-1245, 2008.
 [17] R. Ajemian, et al., "Assessing the function of motor cortex: single-neuron models of how neural response is modulated by limb biomechanics." *Neuron*, vol.58, pp. 414-28, 2008.
 [18] J.P. Scholz and G. Schöner, "The uncontrolled manifold concept: identifying control variables for a functional task." *Exp. Brain Res.*, vol.126, pp. 289-306, 1999.
 [19] A.G. Auyang, J.T. Yen, and Y.H. Chang, "Neuromechanical stabilization of leg length and orientation through interjoint compensation during human hopping." *Exp. Brain Res.*, vol.192, pp. 253-64, 2009.
 [20] J.T. Yen, A.G. Auyang, and Y.H. Chang, "Joint-level kinetic redundancy is exploited to control limb-level forces during human hopping." *Exp. Brain Res.*, To be published.
 [21] C.T. Farley, et al., "Hopping frequency in humans: a test of how springs set stride frequency in bouncing gaits." *J. Appl. Physiol.* vol.71, pp. 2127-32, 1991.
 [22] D.P. Ferris, et al., "Neuromechanical adaptation to hopping with an elastic ankle-foot orthosis." *J. Appl. Physiol.* vol.100, pp. 163-70, 2006.
 [23] D.A. Winter, *Biomechanics and Motor Control of Human Movement*. 2nd ed. 1990, New York: John Wiley & Sons.
 [24] O. Khatib, "A unified approach to motion and force control of robot manipulators: The operational space formulation." *IEEE J. Robot. Automat.* vol.3, pp. 43-53, 1987.
 [25] J.P. Scholz, D. Reisman, and G. Schöner, "Effects of varying task constraints on solutions to joint coordination in a sit-to-stand task." *Exp. Brain Res.* vol.141, pp. 485-500, 2001.
 [26] H. Müller and D. Sternad, "A randomization method for the calculation of covariation in multiple nonlinear relations: illustrated with the example of goal-directed movements." *Biol. Cyb.* vol.89, pp. 22, 2003.
 [27] W.J. Conover, *Practical Nonparametric Statistics*. 1980, John Wiley & Sons, Inc.: Hoboken, NJ.