Joint-Level Responses to Counteract Perturbations Scale with Perturbation Magnitude and Direction

Mark Vlutters, Edwin H.F. van Asseldonk and Herman van der Kooij

Abstract To realize a lower extremity exoskeleton that can provide balance assistance in a natural way, an understanding of human balance control is a necessity. In this study, we investigated how the angle, torque and power of the ankle, knee and hip joints changed in response to balance perturbations during walking. Nine healthy young adults walked on an instrumented treadmill and received pelvis perturbations of various magnitudes and directions at the instance of toe-off right. An open source musculoskeletal modeling package (OpenSim) was used to perform inverse kinematics and inverse dynamics. Subjects modulated the ankle torque in the (left) stance foot with the magnitude and direction of the perturbation. Also in gait phases following foot placement, subjects addressed ankle torques to mitigate the remaining effects of the perturbation. The results presented here support the use of ankle actuation in lower extremity orthoses for natural and cooperative balance control.

1 Introduction

Most current exoskeletons are unable to stay upright without assistance and guidance of its user. Paraplegic users, for example, often require crutches to prevent falling. To have an exoskeleton *assist* its user in maintaining balance instead,

This work was supported by the BALANCE (Balance Augmentation in Locomotion, through Anticipative, Natural and Cooperative control of Exoskeletons) project, partially funded under grant 601003 of the Seventh Framework Program (FP7) of the European Commission (Information and Communication Technologies, ICT-2011.2.1).

M. Vlutters (✉)
Biomechanical Engineering Group, University of Twente, Enschede, Netherlands
e-mail: m.vlutters@utwente.nl

E.H.F. van Asseldonk · H. van der Kooij
University of Twente, Enschede, Netherlands
e-mail: {e.h.f.vanAsseldonk,h.vanderkooij}@utwente.nl

J. González-Vargas et al. (eds.), *Wearable Robotics: Challenges and Trends*,
Biosystems & Biorobotics 16, DOI 10.1007/978-3-319-46532-6_23

preferably in a natural and human-like manner, an understanding of human balance control is of major importance.

Investigating balance responses following perturbations can give insight in the human balance controller. When and how are the different lower extremity joints controlled to maintain upright posture? Furthermore, insight into joint angles, torques, and power during perturbation recovery can provide guidelines for exoskeleton hardware specifications.

In this study, pelvis perturbations were used to elicit balance recovery responses in walking human subjects. We investigated how joint-level responses alter with perturbation magnitude and direction, and provides insight in the ranges of motion, torque, and power of the ankle, knee, and hip joints during the recovery.

2 Materials and Methods

2.1 Experimental Setup and Protocol

Nine healthy young adults walked on a custom dual-belt instrumented treadmill (MotekForce Link, Culemborg, Netherlands), while keeping arms crossed over the abdomen. Walking speeds were 2.25 km/h and 4.50 km/h, scaled to the subject's leg length. For each speed, subjects first walked a 2 min unperturbed baseline trial. In subsequent trials, perturbations were applied to the pelvis using one of two motors (Moog, Nieuw-Vennep, Netherlands), connected with a lever arm to a pelvic brace (Distrac, Hoegaarden, Belgium), worn by the subject. The motors were located at the side and the rear of the treadmill. Perturbations were applied at randomly selected instances of toe-off right (TOR), and consisted of 150 ms block pulses of a force magnitude equal to 4, 8, 12 and 16 % of the subject's body weight. Perturbation directions were forward (positive), backward (negative), outward (positive, away from stance leg) and inward (negative, toward stance leg). Each perturbation type was repeated 8 times, leading to 256 perturbations per subject. Kinematic data of various landmarks on the lower extremities, pelvis, trunk and head [1] were collected at 100 Hz using a motion capture system (Phoenix Technologies Inc, Vancouver, Canada). Ground reaction forces, subject-motor interaction forces, and EMG data were collected at 1000 Hz. More details about the experimental setup can be found in [2].

2.2 Data Processing

Data were processed using Matlab (R2014b, Mathworks, Natick, US) and OpenSim 3.3 [3]. Joint angles and velocities were calculated using inverse kinematics (IK), joint torques using inverse dynamics (ID), and joint power by multiplying the joint velocities and torques. The model used for the IK and ID calculations in OpenSim was the default gait2354 model, scaled to subject specific dimensions using the kinematic landmarks from each subject's static measurement. In the ID, the collected ground reaction forces as well as the subject-motor interaction forces were specified as external force.

All data were cut in sequences from (1) perturbation onset (at TOR) to perturbation end, (2) perturbation end to the subsequent heel strike right (HSR), (3) HSR to subsequent toe-off left (TOL), (4) TOL to subsequent heel strike left (HSL), (5) HSL to subsequent TOR. For each subject, all sequences were resampled to 50 samples and averaged over the repetitions. The repetition averages of each subject were used to obtain averages and standard deviations over subjects.

3 Results

Here, results are only shown for the plantar- and dorsiflexion ankle torque and ankle power, in response to anteroposterior (AP) perturbations for the 2.25 km/h walking speed (Fig. 1). Subjects scaled their left ankle torque with perturbation magnitude and direction during the left single support phase, directly following the perturbation. During this phase subjects extracted energy following forward perturbations, and injected energy following the larger backward perturbations, as can be observed from the power.

Using the right ankle, subjects delivered an increased plantarflexion torque directly following HSR in response to the larger forward perturbations, resulting in increased energy extraction (negative power). In the second double support phase following the perturbation (HSL-TOR), subjects generated more positive power following backward perturbations as compared to forward perturbations. Surprisingly, the perturbation effects are more pronounced in the right ankle during this second double

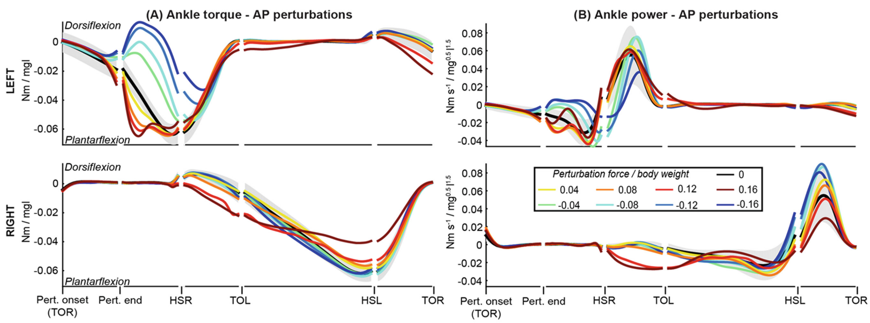

Fig. 1 Ankle plantar- and dorsiflexion torques (A) and power (B) in response to anteroposterior pelvis perturbations during 2.25 km/h walking. Top row: left ankle, bottom row: right ankle. Colors indicate the different perturbation magnitudes. Shaded gray area indicates the baseline standard deviation. Standard deviations of the perturbation data are not shown to prevent image cluttering. Data were made dimensionless using subject weight (m*g) and height (l).

support phase, than in the power delivered by the left ankle during the first double support phase (HSR-TOL) after the perturbation.

4 Discussion

Joint angles, torques, and power were obtained using IK and ID. The ankle torques and power show that an ankle strategy is actively addressed in the recovery from pelvis perturbations during walking. One might expect a decreased plantarflexion torque during the push-off directly following forward perturbations (Pert.end-HSR), but an increase was observed instead. An explanation could be that subjects attempt to keep their center of mass at approximately the same height to prevent having to strongly redirect the body vertically [4], or prevent forward body rotation as in [5]. Both require leg extension through plantarflexion. The strong decrease in left ankle plantarflexion torque directly following the perturbation (Pert.end-HSR) allows subjects to quickly regain forward velocity and return to the desired gait cycle. Consequently, no strong ankle torque deviations are observed in the subsequent gait phases following backward perturbations. The larger variability between conditions in right peak ankle power (HSR-TOL) compared to left peak ankle power (HSL-TOR) might be related to the subject repositioning on the treadmill, which likely does not occur until the second step (HSL).

5 Conclusions

The presented results can give insight in human balance control on a joint level. Future work consists of finding controllers that can generate such joint-level responses.

References

1. Dumas, R., Chèze, L., Verriest, J.P.: Adjustments to McConville et al. and Young et al. body segment inertial parameters. J. Biomech. **40**, 543–553 (2007)
2. Vlutters, M., van Asseldonk, E.H.F., van der Kooij, H.: Center of mass velocity based predictions in balance recovery following pelvis perturbations during human walking. J. Exp. Biol. **209** (2016, to be published)
3. Delp, S.L., et al.: OpenSim: open-source software to create and analyze dynamic simulations of movement. IEEE Trans. Biomed. Eng. **54**, 1940–1950 (2007)
4. Donelan, J.M., Kram, R., Kuo, A.D.: Mechanical work for step-to-step transitions is a major determinant of the metabolic cost of human walking. J. Exp. Biol. **205**, 3717–3727 (2002)
5. Pijnappels, M., Bobbert, M.F., van Dieën, J.H.: How early reactions in the support limb contribute to balance recovery after tripping. J. Biomech. **38**, 627–634 (2005)