OPTIMAL COMBINATION OF MINIMUM DEGREES OF FREEDOM TO BE ACTUATED IN THE LOWER LIMBS TO FACILITATE ARM-FREE PARAPLEGIC STANDING

Joon-Young Kim[1,3], Albert H. Vette[2,3], James K. Mills[1], and Milos R. Popovic[2,3]

[1] *Department of Mechanical and Industrial Engineering, University of Toronto, Canada*
[2] *Institute of Biomaterials and Biomedical Engineering, University of Toronto Canada*
[3] *Toronto Rehab, Lyndhurst Centre, Toronto, Canada*

INTRODUCTION

A paraplegic is an individual whose spinal cord has been injured below the thoracic vertebra No.1, resulting in partial or complete loss of motor function in the lower limbs. The lost ability to stand can be regained by artificially stimulating the skeletal muscles in the lower limbs that cannot be controlled voluntarily. Recently, arm-free standing of individuals with paraplegia using functional electrical stimulation (FES) has drawn much attention in the field as it might allow paraplegics to stand and at the same time use both arms to perform activities of daily living (ADL) [1,2]. However, current FES systems for standing do not open up this possibility yet. Instead, they require that the individual actively regulates balance using one or both arms, thus limiting the use of the FES systems during ADL (e.g., [3]).

Estimating the joint torques in the lower limbs that would allow stable standing is an important first step towards determining the feasibility of FES-assisted arm-free standing of individuals with paraplegia. Once the joint torques needed during arm-free standing are known, the muscles capable of generating these torques can be identified. This would eventually assist in determining viable muscle stimulation strategies that an FES system for arm-free paraplegic standing should apply [4].

In our previous study [5], we developed a dynamic model of the human body during double-support stance and found that only six out of twelve degrees of freedom (DOF) need to be actuated in the lower limbs to facilitate stable arm-free standing despite moderate disturbances. A total of six different cases of six active DOF (6-DOF) have been identified that would facilitate stable arm-free standing. Those six cases of 6-DOF were proposed in order to minimize the number of DOF that must be actuated by a potential FES system for paraplegic arm-free standing. However, it is certain that the six cases of 6-DOF will not generate identical torque patterns since each case consists of a different combination of DOF in the lower limbs.

Accordingly, in this manuscript, the joint torques were estimated using the inverse dynamics method for all six 6-DOF systems. The purpose was to show that actuating only six out of twelve DOF in the lower limbs to allow individuals with paraplegia to stand freely is theoretically feasible with respect to multi-body stability. Specifically, the goal was to determine the *optimal* 6-DOF combination that can be actuated in the lower limbs using FES to ensure stability during perturbed arm-free standing.

METHODS

Three-Dimensional (3D) Dynamic Model

The 3D dynamic model of the human body during double-support stance, which has been developed in our previous study [5], is shown in Fig. 1. The HAT (Head-Arms-Trunk) was modeled as a rigid body with a constant mass and moment of inertia. Each leg consisted of six DOF, which has been shown to be a reasonable approximation of the double-support stance in humans [6]. The feet were not included in the model since it was assumed that they were fully in contact with the ground. Denavit-Hartenberg notation was used for kinematic modeling, and Newton-Euler and Lagrange formulations were applied to the HAT and the legs, respectively, to obtain the dynamic model of the system. Refer to Kim *et al.* [5] for more details.

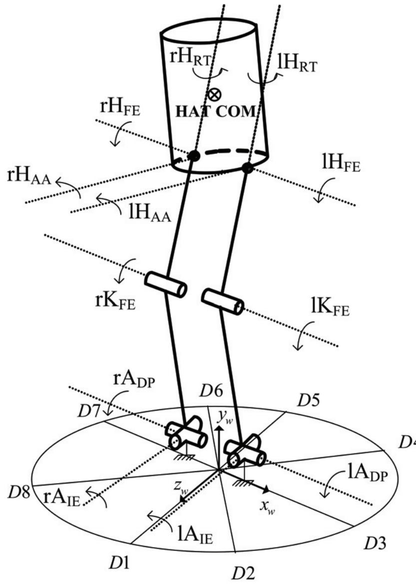


Figure 1: 3D dynamic model of the human body during double-support stance [5]. The DOF of the joints are: ankle in-/eversion (lA_{IE}, rA_{IE}); ankle dorsi-/plantarflexion (lA_{DP}, rA_{DP}); knee flexion/extension (lK_{FE}, rK_{FE}); hip ab-/adduction (lH_{AA}, rH_{AA}); hip flexion/extension (lH_{FE}, rH_{FE}); and hip rotation (lH_{RT}, rH_{RT}). The system was perturbed in eight different directions, D1 to D8.

Inverse Dynamics Solutions

To find the inverse solution of the dynamic model, *Nakamura's method* was applied [7]. As long as the number of the active DOF $\geq$ 6 and the rank of the Jacobian matrix with respect to the passive DOF is equal to 6, the inverse dynamics solution exists [5].

Table 1: Six cases of six active DOF combinations

	Ankle		Knee		Hip			
	rA_{DP}	lA_{DP}	rK_{FE}	lK_{FE}	rH_{AA}	lH_{AA}	rH_{FE}	lH_{FE}
I		•	•	•		•	•	•
II	•		•	•		•	•	•
III	•	•		•		•	•	•
IV	•	•	•			•	•	•
V	•	•	•	•	•		•	
VI	•	•	•	•		•	•	

Table 1 describes six combinations of six active DOF, i.e., the minimum number of DOF, which obtained the full Jacobian rank for all feasible motions during quiet standing. Note that the six combinations of active DOF in Table 1 are based on the assumption that lA_{IE}, rA_{IE}, lH_{RT}, and rH_{RT} were always passive DOF.

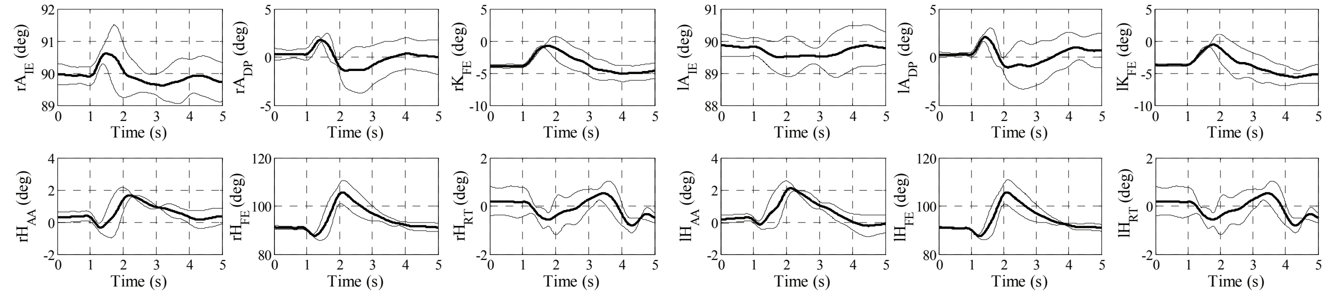
Acquisition of Able-Bodied Dynamics

To estimate the joint torques with the inverse dynamics method, the body kinematics of four healthy individuals during perturbed arm-free standing were obtained. The subjects (age: 31 – 32 years) were asked to stand with bare feet in the most comfortable standing posture with the eyes closed and the arms folded on the chest. In order to obtain an accurate 3D representation of the kinematics during standing, the subjects were perturbed using an impulse-like force acting in eight directions (D1–D8 in Fig. 1). The perturbation system consisted of a drop weight of 4.54 kg, a pulley and a rope that was attached to each subject at the level of his/her center of mass (COM). For each perturbation direction, the rope connected to the waist of the subject was cocked (zero slack), suddenly pulled by releasing the drop weight, and then slackened in order to generate impulse-like perturbations (three times for each direction).

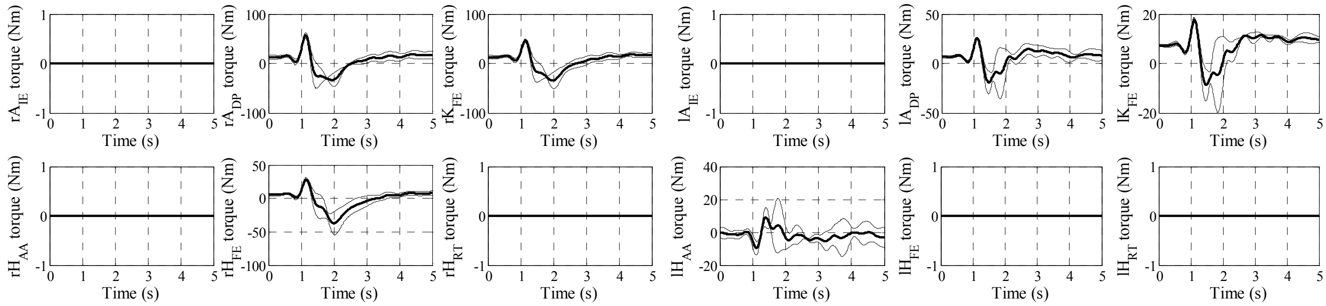
A motion analysis system (Optotrak 3020, Northern Digital, Canada) recorded the joint angles of the body segments with a sampling frequency of 100 Hz. The joint angles were filtered using a fourth-order, zero-lag Butterworth filter with a cut-off frequency of 2 Hz. The angular velocities and accelerations were then estimated by the *finite differentiation technique* [4]. The lengths of the body segments of each subject were measured before the experiments, whereas the mass, the moment of inertia, and the location of the COM of each body segment were estimated based on the subject's anthropometric data [4]. By means of the kinematic data and the anthropometric parameters of the subjects, the joint torques of the minimum number of DOF were calculated for all eight perturbation directions and all six cases shown in Table 1.

Analysis of Joint Torques

The analysis of the joint torques was performed to identify how each case of 6-DOF



(a) Joint angles



(b) Joint torques

Figure 2: Joint angles (a) and joint torques at the six actuated DOF of Case VI (b) due to the perturbation given in direction D1 (subject S1). The mean values with one standard deviation from three trials are shown as a function of time.

shown in Table 1 was different from the other five cases. A norm based measure for each active DOF's torque profile was determined by numerically integrating the norm of the torque as a function of time over an interval of five seconds, i.e., $\int_0^5 |\boldsymbol{\tau}_a| dt$ (Nm·s), where $\boldsymbol{\tau}_a$ is the active DOF's mean torque as a function of time obtained from the three perturbation experiments per subject. The value of $\int_0^5 |\boldsymbol{\tau}_a| dt$ is referred to in the paper as *torque sum*.

RESULTS

The inverse dynamics method revealed that the recorded body kinematics could be generated by feasible joint torques for all six cases of six active DOF. Examples of the recorded body kinematics and the calculated joint torques for Case VI (D1 perturbation of subject S1) are shown in Fig. 2.

In Fig. 3, the overall torque sums at all active DOF calculated for all eight directions of perturbation and all six 6-DOF combinations (Table 1) are presented. Cases V and VI generated the lowest overall torque sum at all six active DOF for all directions of perturbation. Cases I and II generated the largest overall torque sum at all active DOF for the forward perturbations (D1, D2, and D8), whereas Cases III and IV generated the largest overall torque sum at all active DOF for the backward perturbations (D4, D5, and D6).

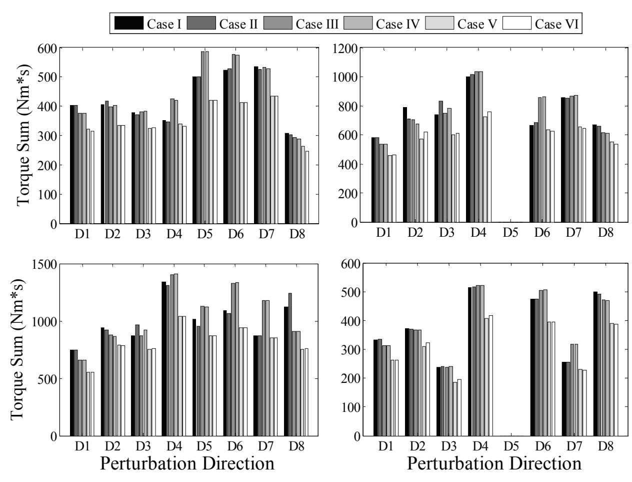


Figure 3. Overall torque sums at all active DOF, shown for each of the six 6-DOF systems (Cases I to VI) and all directions of perturbation (D1 to D8). Note that subjects S2 and S4 lost balance during the backward perturbations (D5).

DISCUSSION

The most significant result obtained from the joint torque estimation is that there exist optimal cases among the 6-DOF combinations such that the overall torque sum was lowest and most evenly distributed among the two legs for all eight perturbation directions. The optimal combinations consisted of one H_{AA}, one H_{FE}, two K_{FE}, and two A_{DP} (Cases V and VI). Note that these two cases actually imply four different optimal combinations as each case in Table 1 has a counterpart due to the symmetry of the two legs. Cases I to IV on the other hand had two H_{FE}, thus missing one DOF to be actuated in either the ankle or knee joint. The joint torque patterns of Cases V and VI were fundamentally different from the other cases in that the torque of all the active anterior/posterior DOF acted together in the same direction, whereas in the other cases at least one DOF existed that was actuated in the opposite direction due to the closed-chain dynamics (not shown). As such, it seems plausible that Cases V and VI require less amount of torque compared to the other four cases.

Our theoretical analysis represents a crucial first step towards developing an FES system for arm-free standing. It implies that the joint torques needed to stabilize the human body during perturbed standing can be minimized with the determined optimal combination of active DOF, and that four variations of this combination exist from which an appropriate set can be chosen for a particular individual with paraplegia. However, many other challenges need yet to be addressed. These include the development of a balance sensor for sensory feedback; the compensation of FES induced muscle fatigue; and the identification of an appropriate closed-loop control strategy to regulate FES induced muscle contractions.

CONCLUSION

In conclusion, FES-assisted, arm-free standing is theoretically feasible with respect to multi-body stability. This is true even when only six out of twelve DOF in the lower limbs are actuated. Moreover, an *optimal* combination of 6-DOF exists that minimizes the energy spent due to the FES-elicited muscle contractions. Our findings are especially relevant for individuals with paraplegia who are potential users of FES-assisted standing technology, since the muscles actuating specific DOF are often denervated or difficult to access. Additionally, it is preferred to develop an FES system with less stimulation electrodes (i.e., an FES system for 6 DOF rather than for 12 DOF) in order to reduce system complexity and cost, but increase the system's reliability.

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