Assisting gait with free moments or joint moments on the swing leg

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Abstract—**Wearable actuators in lower-extremity active orthoses or prostheses have the potential to address a variety of gait disorders. However, whenever conventional joint actuators exert moments on specific limbs, they must simultaneously impose opposing reaction moments on other limbs, which may reduce the desired effects and perturb posture. Momentum exchange actuators exert free moments on individual limbs, potentially overcoming or mitigating these issues.**

We simulate unperturbed gait to compare conventional joint actuators placed on the knee or hip of the swing leg, and equivalent angular momentum exchange actuators placed on the shank or thigh. Our results indicate that, while conventional joint actuators excel at increasing toe clearance when assisting knee flexion, free moments can yield greater increases in stride length when assisting knee extension or hip flexion.

I. INTRODUCTION

biological joints and actuating in parallel with the musculature by exerting opposite reaction moments on the adjacent limbs. However, these opposite moments do not directly contribute to a net change in angular momentum and risk internally perturbing posture.

Angular momentum exchange actuators (AMEAs), such as reaction wheels and control moment gyroscopes, provide exciting new possibilities for wearable robotics. Unlike conventional actuators, which exert opposing moments between two bodies connected by a joint (joint moment, JM, Fig. 1a), AMEAs exert moments between a body and a rotating mass contained within the actuator, where the result is similar to a free moment (FM) or moment exerted against an inertially-fixed body. For a wearable device, this entails that (i) the actuator need not be placed on a joint, but at [illegible] location on a body segment, (ii) a net contribution to angular momentum can be made [illegible], and (iii) no opposite reaction moments are exerted by the actuator on adjacent body segments, reducing the risk of internal perturbation. This would enable, for example, a transfemoral prothesis containing an AMEA to provide assistance to the hip, even without a structure spanning the hip, which could benefit amputees exhibiting gait asymmetry due to muscle atrophy around the residual joint [24].

Wearable AMEAs comprising reaction wheels or control moment gyroscopes have been described in backpack-like balance aids [25]–[28], while others have envisaged them placed on the limbs for either emulation of a viscous environment [29], actuating or replicating lost function in upper extremity prostheses [30], [31], or assisting knee and hip placement of this mass at different locations on the body are left for future investigation.

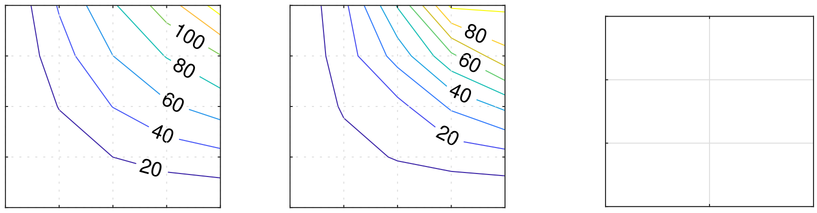
All moment profiles were parameterized as a rectangular profile beginning at time t_s after toe-off of the right leg, with magnitude M and duration Δt (Fig. 1b) – this shape was taken for convenience and is not claimed to be optimal. The parameters were discretized into 3D grids for analysis, with $M \in [5, 30]$N m (5 N m increments) intended to represent realistic capabilities of a wearable actuator, and $\Delta t \in [50, 300]$ms (50 ms increments) was selected to span the majority of the swing phase (approx. 530 ms). For assisting knee flexion, moments were applied only in the early-to-mid swing phase (approx. 0-200 ms), so Δt was truncated to $[50, 200]$ms and the start time selected as $t_s \in [60, 180]$ms (20 ms increments). Knee extension was performed in mid-to-late swing phase and hip flexion in early-to-late swing phase, so $t_s \in [180, 290]$ms (20 ms increments) and $t_s \in [120, 240]$ms (20 ms increments) were selected, respectively.

Changes in stride length (measured between successive heel strikes of the same foot) and minimum toe clearance (MTC) of the actuated swing leg were selected as the primary outcome measures for comparing JM and FM in each application (KF, KE, HF). For the purposes of this analysis, the parameter t_s is not of interest, so was selected to maximize either MTC (KF, HF) or stride length (KE, HF) for each actuator type and each combination of parameters M and Δt. To prevent artifacts (e.g. a null-space) in the non-optimized outcome measure, t_s was computed as:

$$t_s = \arg\max \left(\lambda \, \mathrm{MTC} + (1 - \lambda) \mathrm{SL} \right) \ , \tag{1}$$

where $\lambda = 0.99$ to (primarily) maximize MTC or $\lambda = 0.01$ to maximize stride length (SL).

Quantification of differences between actuators and applica-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