USING MUSCULOSKELETAL MODELS TO ESTIMATE THE PASSIVE JOINT STIFFNESS

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INTRODUCTION

Joint stiffness is actively modulated by healthy humans during tasks requiring interaction with the environment [1]. For individuals with neuromotor dysfunctions, however, the response of the neuromuscular system to exogenous perturbations is compromised [2]. As such, analysis of joint stiffness during manipulation could be important both for basic research in motor neuroscience, and for clinical assessment purposes.

Joint stiffness can be described by two different components: active and passive. The active component is related to the force generating capability of muscles; the passive component is an intrinsic property that depends on the passive elastic behavior of the muscles.

Although experimental approaches to the computation of the passive joint stiffness have been presented [3], the complexity and the length of the experiments limit the number of postures in which joint stiffness can be computed. Musculoskeletal models (MSMs) could extend the results of human subject experiments, and estimate the stiffness of different joints across their entire workspace. However, MSMs have not been developed for joint stiffness analysis; even the simplest models can be characterized by unstable values of stiffness that are not physiologically accurate [4].

This work develops a framework to analyze joint stiffness in MSMs and presents a procedure to minimally modify some of the muscles parameters of a MSM to obtain stable and physiologically accurate values of passive joint stiffness.

METHODS

In static conditions, joint stiffness (K_θ) is the derivative of the multivariate torque (τ) vs. angle (θ) relationship:

$$K_\theta = \frac{\partial \tau}{\partial \theta} \qquad (1)$$

where τ is a function of the musculotendinous forces (f^{MT}) and the muscle Jacobian (J) (i.e. $\tau = J^T f^{MT}$). Assuming that muscles are modeled with a Hill-type model, and considering only the muscle's passive component, it is possible to express f^{MT} as:

$$f^{MT} = f^T = f^M = f^{PE}(l^{PE})\cos(\alpha) \qquad (2)$$

where f^T and f^M are the forces along the tendon and muscle respectively; α is the pennation angle.

It is possible to show that:

$$K_\theta = J^T \frac{1}{c}\frac{K^{PE}K^T}{K^{PE}+K^T} J - \frac{\partial J^T}{\partial \theta} f cos(\alpha) + J^T f \sin(\alpha)\frac{\partial \alpha}{\partial \theta} \qquad (3)$$

where $c = \left(1 + \frac{l^{PE}}{\partial l^{PE}}\frac{\partial \cos(\alpha)}{\cos(\alpha)}\right)$. We consider a joint unstable if at least one of the eigenvalues of K_θ is negative.

Since the first term of (3) is always positive, negative (i.e. unstable) values of stiffness can only be introduced by the second and third terms. The parameters that influence stability are then the moment arm (J), its derivative ($\frac{\partial J}{\partial \theta}$), the muscle force ($f^{PE}$), the pennation angle ($\alpha$) and its derivative $\left(\frac{\partial \alpha}{\partial \theta}\right)$. Preliminary analysis shows that the third term of (3) influences the final value of stiffness by only 0.1%. As such, we consider the moment arm and the muscle force as the only relevant parameters.

We applied the described analysis to study the stiffness of the wrist joint implemented in the upper limb model proposed by Holzbaur et al. [5]. We used a simplified version of the model where only five muscles (ECRL, ECRB, ECU, FCR, FCU) were considered. We calculated stiffness in a 2D workspace defined by two wrist rotations, flexion/extension (FE) with a range of -50° (extension) to 50° (flexion), and radial/ulnar deviation (RUD), ranging from -25° (radial deviation) to 10° (ulnar deviation).

Since the model is unstable in certain regions of the workspace (see results), we developed a framework to determine how to minimally modify the model to obtain stability. As both the moment arm [6] and the tendon slack length [7] are highly variable between individuals and cannot be measured accurately, we hypothesized that that small modifications to these parameters could yield equally accurate MSMs, which are also stable and suitable for the analysis of dynamic interaction with the environment.

We conducted an exhaustive search on the space defined by each muscle attachment point and tendon slack length, and used optimization to find a set of parameters that achieves stability in all workspace postures while minimally changing model behavior. Our analysis seeks to minimize the following function:

$$\Phi = a_1 \frac{A_{unst}}{A_{tot}} + a_2 \frac{\Delta L_{ts}}{\Delta L_{ts_{MAX}}} + a_3 \frac{\Delta \rho_{IP}}{\Delta \rho_{IP_{MAX}}} + a_4 \frac{A_{sl} - A_{sl_0}}{A_{tot}},$$

where A_{unst} and A_{sl} are the area of the unstable portion and of the slack portion of the workspace, respectively; ΔL_{ts} is the tendon slack length variation; $\Delta \rho_{IP}$ is the insertion point displacement.

A preliminary analysis showed that, by keeping the flexors unloaded in the entire workspace, as proposed by [5] and shown in Fig. 1A, the model presents a partial area where all the muscles are slack. Considering this a non-physiological condition, we forced to the FCR and FCU to decrease the slack area by 20% (i.e. the same amount of modification required to stabilize the ECRL and ECRB).

RESULTS

Model stability analysis:

We analyzed joint stability separately for each muscle of the model, and created the stability maps reported in Fig. 1A, which demonstrated the presence of a large unstable region in the workspace. The instability was caused by two muscles (ECRL and ECRB) that, if stretched, are always unstable.

Stabilization of unstable muscles:

With the optimization procedure, the attachment points and tendon slack length of the unstable muscles were modified to achieve stability throughout the workspace for each muscle and for the entire model (Fig. 1B). Muscle moment arms for the stabilized muscles were changed substantially only for ECRB in the RUD axis, for which the average change in moment arm required for model stabilization was 280%.

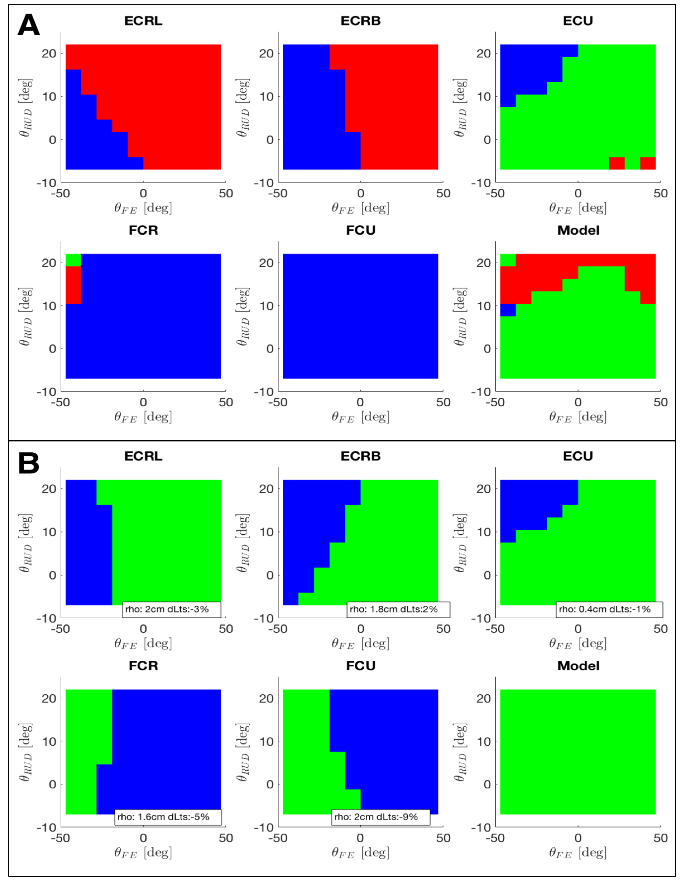

Figure 1: Stability maps for each muscle individually and for the entire model: (A) original model based on [5], (B) optimized model. In each map the green area is the stable area, the red area is the unstable area and the blue area is the slack area.

CONCLUSIONS

Using a musculoskeletal model to investigate the joint stiffness is valuable, as it may allow for the exhaustive analysis of the joint stiffness properties.

Even though MSMs have not been developed to analyze the joint stiffness and can show unstable behavior, we developed a novel method that allows stability to be achieved through minimal parameter changes of the MSM. With proper experimental validation, this method may allow to use current MSMs to study joint stiffness to optimize human-robot interaction and/or to quantify the evolution of disease in neuromotor disorders.

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