Lower-Limb Multi-Joint Stiffness of Knee and Ankle

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***Abstract*— Lower-limb multi-joint (knee and ankle) stiffness may play an important role in functional activities such as walking, and may be significantly altered post stroke. Thus, determination of lower-limb multi joint stiffness matrix is important for better understanding of gait and of pathological changes post stroke. In this study, using novel dynamics decomposition, the knee and ankle joint stiffness matrix including cross-coupled stiffness terms between the two joints were determined and reported ever first. The determined stiffness matrix may be useful for gait studies, and can be served as a baseline for studying pathophysiological changes post stroke.**

I. Introduction

Hypertonia, involving spasticity and/or contracture, is a major source of disability following stroke, and it impairs the control of knee and ankle substantially [1-3]. The hypertonus and reflex hyperexcitability disrupt the remaining functional use of muscles, impede motion, and may cause severe pain and structural changes of muscle fibers and connective tissue [4], [5], which may result in alterations of intrinsic mechanical properties (and reduction of range of motion (ROM)) of joints involved [6], and lead to a clinical contracture [1], [2], [7], [8]. Alteration in muscle [5] and tendon [9] properties can modify overall intrinsic mechanical properties of *multiple joints* of impaired lower-limb [6] that can potentially reduce ROM and may lead to a clinical contracture [1], [2], [7], [8] with potential stereotypical lower-limb deformity patterns [4], [5], [10]; and can induce changes in multiple components of lower-limb multi-joint stiffness matrix (resistance to passive movement) including inter-joint cross-coupled terms, which is empirically well-known but difficult to identify through clinical manual examinations [11-19]. The individual joint stiffness terms (the diagonal terms) include mono- and bi-articular muscles' contributions and the cross-coupled stiffness terms (the off-diagonal terms) include bi-articular muscles' contributions (e.g., Gastrocnemius (GAS)) [15]. Alteration in lower-limb muscles (and tendons) is also associated with stroke survivors' impaired gait (e.g., foot-drop [20-23], knee-hyperextension [22-24], and coupling of the ankle and knee impairments [25], [26]). Thus, identifying changes in the stiffness matrix post-stroke may help us better understand stroke survivors' impaired gait and develop more effective stroke rehabilitation. Despite of its importance, even for healthy individuals, there has been a lack of simultaneous characterizations of knee and ankle stiffness matrix, including the cross-coupled stiffness, though the knee and ankle joint stiffness was characterized individually [27], [28]. In this regard, there are strong scientific and clinical needs to characterize multi-joint coupled mechanical stiffness matrix of knee and ankle dynamics and to quantify changes of the multi-joints dynamics post-stroke simultaneously using a robust identification method. In this paper, as a first step, healthy subjects' stiffness matrix was determined.

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II. Materials and Methods

A. Experimental Setup

A custom 2 degree-of-freedom exoskeleton-robot-type device driving knee and ankle joints perturbed the two joints and recorded the two joints' torques and angles (Fig. 1). The subject was seated with the thigh and the trunk strapped to the seat and shank attached to the long link of the device through a rigid brace. The rotation axis of knee motor was aligned with that of the knee flexion. The ankle motor could be shifted along the long link to align the rotation axis of the motor with that of the ankle dorsiflexion. Torques and angles of knee and ankle were measured from torque sensors and encoders at the two joints. With immobilization of the ankle by the ankle motor isometrically, the knee joint was moved at a slow constant speed by the knee motor and *vice versa*, while the subject remained relaxed. The two motors were precisely controlled with readings of positions, velocities, and torques of the two joints during the experiment. A stop switch was given both to the operator and to the subject to authorize them to shut down the whole system at any time. Torques and angles of the knee-ankle joints were recorded at 1 KHz during each trial. The knee and ankle were perturbed at 9 positions

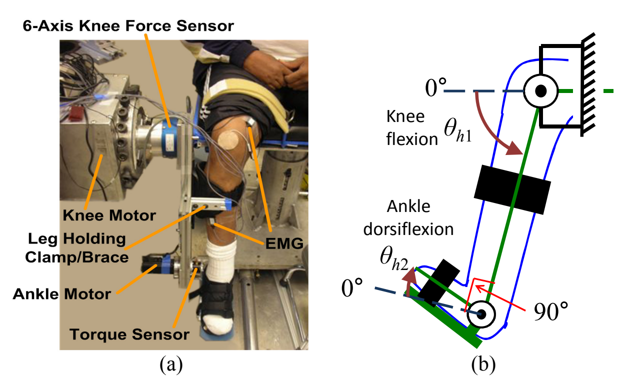

Fig. 1. (a) Multi-joint setup for studying knee-ankle dynamics. (b) Sagittal plane schematic diagram. The subject was seated with the thigh and trunk strapped to the seat and shank attached to the link supporting the shank through a rigid brace.

consisting of combinations of three knee positions (15º, 45º and 90º flexion to separate GAS contribution to knee and ankle joint torques [4], [29-31]) and three ankle positions (15º, 0º and -30º dorsiflexion), to cover the functional ROMs of the two joints [5], [32], and to evaluate the stiffness matrix at different joint positions considering the stiffness may vary systematically with the two joints' positions [4], [5], [11], [29], [33] and so does the hypertonic reflex [34]. For all conditions, hip was at 85º flexion.

B. Multi-Input Multi-Output Modeling and Identification of Multi-Joint Coupled System

Human limbs may behave linearly [11], [16-18], [35], [36]. At around an fixed operating state[1] λ and at around an equilibrium point of the knee and ankle [16], [17], [37], $\boldsymbol{\theta}_o=[\theta_{h1o}\ \theta_{h2o}]^T$, the multi-joint coupled knee-ankle dynamics can be written as

$$\begin{bmatrix}\tau_{h1}\\ \tau_{h2}\end{bmatrix} = \begin{bmatrix}Z_{h11}(s) & Z_{h12}(s)\\ Z_{h21}(s) & Z_{h22}(s)\end{bmatrix}\left(\begin{bmatrix}\theta_{h1}\\ \theta_{h2}\end{bmatrix} - \begin{bmatrix}\theta_{h1o}\\ \theta_{h2o}\end{bmatrix}\right), \quad (1)$$

where τ_{h1} and τ_{h2} denote knee and ankle torques, respectively, and are elements of vector $\boldsymbol{\tau}_h$; θ_{h1} and θ_{h2} denote knee and ankle angles, respectively, and are elements of vector $\boldsymbol{\theta}_h$; s denotes the Laplace variable; Z_{h11} and Z_{h22} represent knee and ankle local individual joint impedances[2], respectively; Z_{h12} inter-joint cross-coupled impedance relating knee torque (τ_{h1}) and ankle angle (θ_{h2}); Z_{h21} inter-joint cross-coupled impedance relating ankle torque (τ_{h2}) and knee angle (θ_{h1}). Without loss of generality, $\boldsymbol{\theta}_o$ can be set to a zero vector (i.e., $\boldsymbol{\theta}_o=[0\ 0]^T$). Thus, the knee-ankle dynamic in (1) can be rewritten as

$$\begin{bmatrix}\tau_{h1}\\ \tau_{h2}\end{bmatrix} = \begin{bmatrix}Z_{h11}(s) & Z_{h12}(s)\\ Z_{h21}(s) & Z_{h22}(s)\end{bmatrix}\begin{bmatrix}\theta_{h1}\\ \theta_{h2}\end{bmatrix}. \quad (2)$$

For brevity, the Laplace variable s will be omitted in the following discussion. The multi-joint knee-ankle dynamics in (2) will be decomposed, perturbed, and identified using the proposed identification method described in next Subsections.

C. Decomposition of the Multi-Joint Coupled Dynamics

Considering joint stiffness is often changed post stroke, we can focus on the stiffness ($\mathbf{K}_h$) matrix by moving each joint at a slow constant speed to minimize the effect of acceleration related terms and reflex contributions [13], [18], and measuring torques and angles of knee and ankle joints altogether. At a constant speed condition, multi-joint coupled knee-ankle dynamics in (2) becomes

$$\begin{bmatrix}\tau_{h1}\\ \tau_{h2}\end{bmatrix} = \begin{bmatrix}K_{h11} & K_{h12}\\ K_{h21} & K_{h22}\end{bmatrix}\begin{bmatrix}\theta_{h1}\\ \theta_{h2}\end{bmatrix}, \quad (3)$$

where K_{h11}, K_{h22} denote local individual knee and ankle joint stiffness, respectively; K_{h12} inter-joint coupled stiffness between knee torque (τ_{h1}) and ankle angle (θ_{h2}); K_{h21} inter-joint coupled stiffness between ankle torque (τ_{h2}) and knee angle (θ_{h1}).

The decomposition procedure can be applied to (3), to simplify it as follows:

1. The knee is moved at a slow constant speed while the ankle is immobilized by the device. Then, (3) becomes decomposed Single-Input Single-Output (SISO) equations as follows:

$$\tau_{h1} = K_{h11}\theta_{h1} \text{ and} \quad (4)$$

$$\tau_{h2} = K_{h21}\theta_{h1}. \quad (5)$$

Thus, from slow constant speed data, K_{h11} and K_{h21} can be obtained using (4) and (5).

2. Similarly, the ankle is moved at a constant speed while the knee is immobilized by the device. Then, (3) becomes decomposed SISO equations as follows:

$$\tau_{h1} = K_{h12}\theta_{h2} \text{ and} \quad (6)$$

$$\tau_{h2} = K_{h22}\theta_{h2}. \quad (7)$$

Thus, similarly, K_{h12} and K_{h22} can be first obtained from slow constant speed data using (6) and (7).

Clearly, the multi-joint coupled equation is decomposed into four SISO equations. For instance, K_{h11} can be identified from the slope of the knee joint angle-torque curve (i.e., $\partial\tau_{h1}/\partial\theta_{h1}$). In this paper, a simple linear regression was used to obtain the slope of the curve. If needed, for the computation of slopes, either i) a spline with generalized-cross-validation (GCV) [38] or ii) smooth exponential type functions, which has been used for single joint angle-torque curve fitting [13], [39-42], can be used to fit the aforementioned angle-torque curves. Owing to the powerful decomposition, clearly, all the methods used for single joint stiffness identification can be utilized for multi-joint stiffness matrix identification without any major modification.

III. Results

A. Subjects

Two healthy middle age male subjects, who have no previous history of neurological disorder and lower-limb musculoskeletal disease/injury, were recruited for this study. The study was approved by the institutional review board of Northwestern University. A written consent form was obtained from each participant.

B. Experimental Procedure

Each subject's trunk and thigh were strapped to the seat, and shank and foot were attached to the long link of the robotic device using a rigid brace (Fig. 1) aligning with knee flexion/ extension axis and ankle dorsi/planar-flexion axis. Subjects were asked to relax while the device moved their knee and ankle. Their knee was first moved at a slow constant speed ~5 times with ankle immobilized at three different angles (i.e., 15, 0, -30 degree dorsiflexion). Their ankle was then moved at a slow constant moved ~5 times with knee immobilized at three different angles (i.e., 15, 45, and 90 degree flexion). Both joints' angle and torque were saved simultaneously throughout the procedure.

The saved torque and angle data were analyzed using (4), (5), (6), and (7) to obtain the stiffness matrix.

C. Experimental Results

Lower-limb multi-joint (knee and ankle) stiffness matrix was obtained (Fig. 2) at several knee and ankle angles. Each

[1] λ includes mean joint angles, perturbation bandwidth and amplitude, and mean background muscle torque [11], [18].

[2] Impedance ($Z_{hij}(s)$) includes inertia (I_{hij}), viscosity (B_{hij}), and stiffness (K_{hij}): $Z_{hij}(s)=I_{hij}s^2+B_{hij}s+K_{hij}$ (i,j=1,2).

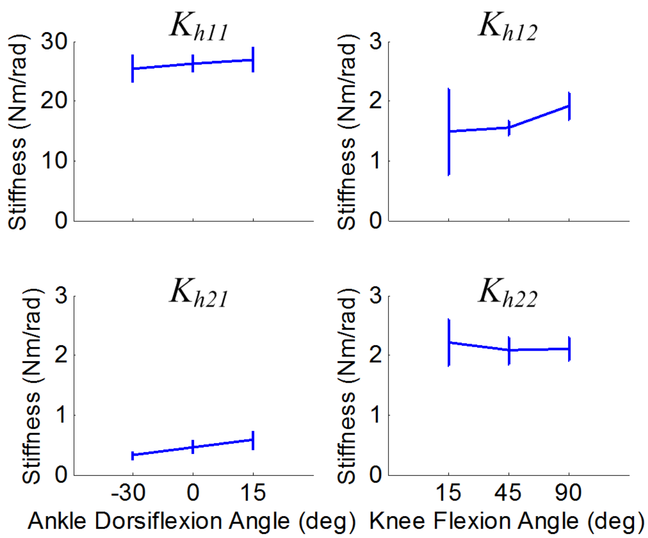

Fig. 2 Multi-joint stiffness matrix at several different knee and ankle angles.

element value changes with knee and ankle angle, indicating possible postural dependence of the stiffness matrix. Second, the two off-diagonal elements (i.e., K_{h21} and K_{h12}) are different from each other in terms of magnitude. In other words, the multi-joint stiffness matrix is not symmetric. Further, the knee individual joint stiffness (i.e., K_{h11}) was an order larger than other three elements.

IV. Discussion and Conclusion

For the first time, knee and ankle multi-joint lower-limb stiffness matrix was determined quantitatively at different knee and ankle angles. The comparison of off-diagonal stiffness terms at 15 degree knee flexion and 90 degree knee flexion may provide the contribution of GAS (a bi-articular muscle) to the stiffness matrix. It seems that with the increase of knee flexion the off-diagonal term (K_{h12}) may increase. Similarly, with increase of dorsiflexion, K_{h11} and K_{h21} may increase. In this study, the two joints' torque measured included gravitational torques due to the foot and shank mass. Thus, the stiffness matrix may include the effect of gravity. Depending on the purpose of identification, sometimes the exclusion of gravity effect may be necessary, but other times may be not. If the purpose of study is to know the stiffness matrix without the effect of gravity (i.e., stiffness matrix mostly due to the muscles and connective tissues), the gravity terms may better be treated separately. If the purpose is more geared toward to know what the resulting effective stiffness in sagittal plane is, for instance, during walking, the gravity effect may be able to be included in the stiffness matrix.

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