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A Multiple Degree of Freedom Lower Extremity Isometric Device to Simultaneously Quantify Hip, Knee and Ankle Torques

Natalia Sánchez [Member, IEEE],

Departments of Biomedical Engineering and Physical Therapy and Human Movement Sciences, Northwestern University, Chicago, IL 60611 USA

Ana Maria Acosta [Member, IEEE],

Department of Physical Therapy and Human Movement Sciences and the Interdepartmental Neuroscience Program (NUIN), Northwestern University, Chicago, IL 60611 USA

Arno H.A. Stienen [Member, IEEE], and

University of Twente Department of Biomechanical Engineering, Enschede, NL and the Department of Physical Therapy and Human Movement Sciences, Northwestern University, Chicago, IL 60611 USA

Julius P.A. Dewald [Member, IEEE]

Departments of Physical Therapy and Human Movement Sciences, Biomedical Engineering, Physical Medicine and Rehabilitation, and Interdepartmental Neuroscience Program, Northwestern University, Chicago, IL 60611 USA

Natalia Sánchez: natalia-sanchez@northwestern.edu; Ana Maria Acosta: a-acosta@northwestern.edu; Arno H.A. Stienen: arnostienen@gmail.com; Julius P.A. Dewald: j-dewald@northwestern.edu

Abstract

Characterization of the joint torque coupling strategies used in the lower extremity to generate maximal and submaximal levels of torque at either the hip, knee or ankle is lacking. Currently, there are no available isometric devices that quantify all concurrent joint torques in the hip, knee and ankle of a single leg during maximum voluntary torque generation. Thus, joint-torque coupling strategies in the hip, knee and concurrent torques at ankle and/or coupling patterns at the hip and knee driven by the ankle have yet to be quantified. This manuscript describes the design, implementation and validation of a multiple degree of freedom, lower extremity isometric device (the MultiLEIT) that accurately quantifies simultaneous torques at the hip, knee and ankle. The system was mechanically validated and then implemented with two healthy control individuals and two post-stroke individuals to test usability and patient acceptance. Data indicated different joint torque coupling strategies used by both healthy individuals. In contrast, data showed the same torque coupling patterns in both post-stroke individuals, comparable to those described in the clinic. Successful implementation of the MultiLEIT can contribute to the understanding of the underlying mechanisms responsible for abnormal movement patterns and aid in the design of therapeutic interventions.

Index Terms

Coupling; Joint Torques; Lower Extremity; Stroke

I. Introduction

Ambulation limitations are common in close to 30 million Americans² who experience chronic conditions such as stroke, cerebral palsy, traumatic brain injury, arthritis or ligamentous injury. Most current methods used to quantify these limitations isometrically are based on measurements of joint kinetics in a single direction using one degree-of-freedom (DOF) sensors such as those in isokinetic dynamometric systems, attached to rigid structures^{3–11}, or quantifications using hand held dynamometers¹³, which should be limited to populations where force/torque magnitudes are small enough to be effectively resisted by hand such as in pediatrics. Voluntary torques generated by the lower extremity are in fact not limited to a single DOF, but are usually combined with torques at other joints, because of the action of biarticular muscles responsible for mechanical coupling between joint torques^{14, 15} and simultaneous neural coactivation of multiple muscles¹⁶. Therefore, quantification of strength in one DOF may not provide enough information about the forces and torques generated by the lower extremity either in healthy individuals or after changes in descending neural drive or changes in musculoskeletal properties seen in muscle and joint diseases.

The use of multi-DOF devices allows for simultaneous quantification of torques generated at the lower extremity joints from hip to ankle: hip abduction/adduction, flexion/extension and external/internal rotation, knee flexion/extension, external/internal rotation and varus/valgus, and ankle dorsi/plantarflexion, and inversion/eversion. Previous studies have aimed to quantify multi-DOF torques in the lower extremity^{17–20}. Specifically, the use of the instrumented Lokomat (Hocoma, Switzerland)^{17, 21} has provided a novel approach to quantify joint torque couples between hip and knee and is the motivation for the development of our device. Studies using the Lokomat provided evidence of abnormal muscle coactivation patterns and joint-torque coupling possibly due to altered supraspinal descending drive^{20, 22–29}. Additionally, studies of isometric lower extremity joint-torque coupling in cerebral palsy quantified stereotypical spontaneous coupling between hip and knee extension²⁰. However, findings with regards to the expression of abnormal joint coupling patterns on individuals post-stroke have not been conclusive and have not obtained consistent results; some studies demonstrated spontaneous coupling of hip adduction with hip and knee extension torques^{17, 20, 21}, coincident with what is described clinically as an “extension synergy” (coupling of hip extension/adduction with knee extension and ankle plantar-flexion³⁰); in contrast, other studies found weakness in specific DOFs but no stereotypical joint torque coupling patterns across subjects^{18, 19}.

A review of the methods used in previous isometric studies^{17–20} indicated differences in the experimental setup that may have influenced torque generation by affecting sensory input, which can alter descending drive^{31–33}. Specifically, studies were performed in recumbent seating²⁰ or standing^{17–19} affecting vestibular and cutaneous input as concluded by Thelen and colleagues²⁰. Also, the amount of body weight support provided differed between

studies from zero^{1, 18, 19} to 100%¹⁷ changing the gravitational contribution to joint torques. Additionally, results could be affected by poorly constrained systems that do not limit movement during maximum torque generation^{15, 18, 19}. Therefore, subjects were instructed to generate maximum voluntary torques while avoiding movement of the poorly constrained limb segments¹⁹, affecting spontaneous coupling across joints. Additionally, the behavior of the non-tested leg was not controlled in any of these studies.

On the other hand, joint torque coupling strategies after peripheral nerve injuries have not been previously investigated. A review of the literature on isometric joint-torque coupling strategies due to peripheral neural, muscle or joint injuries/joint degeneration did not yield any results. Research has mainly focused on quantification of deficits in muscle strength and consequent changes in gait kinematics in multiple sclerosis, knee osteoarthritis and anterior cruciate ligament injury^{9, 34, 35}. Information is lacking on how muscle coactivation patterns and subsequent joint-torque coupling patterns may change in these pathologies to account for muscle weakness or joint pain. The use of multi-DOF devices can aid in the study of the underlying mechanisms responsible for movement abnormalities following peripheral injuries. Once the abnormal joint-torque coupling patterns have been identified, this knowledge can aid in the development of rehabilitation interventions that modify muscle activation strategies and resulting joint-torque coupling.

The present manuscript describes the design, validation and effective implementation of a new Multi-joint Lower Extremity Isometric Torque measurement device, the MultiLEIT. This system can accurately quantify lower extremity joint torque coupling strategies and the distribution of lower extremity joint-torque weakness during generation of voluntary maximal and submaximal torques at the hip, knee and ankle. Also, the materials that compose the MultiLEIT exceed the rigidity of any other systems used previously, eliminating system movement during maximal conditions. First, we will describe the design requirements to guarantee the accuracy of the measurements, comfort and safety. The methods section describes the protocols used to validate and implement the device in a healthy individual and in an individual post-stroke.

We will present validation data and the results from the experimental protocol, demonstrating the capabilities of the device in measuring multi-joint coupling patterns that differ between healthy and post-stroke individuals. Results from studies using the MultiLEIT are expected to aid in the design of improved assistive and therapeutic devices and associated interventions that can be corrective, preventive and targeted for lower extremity rehabilitation of multiple pathologies affecting lower extremity movement.

II. Design

The MultiLEIT measures all isometric torques from hip to ankle in a single leg, with the intention to quantify inter- and intra joint torque couplings in individuals after stroke. Combined with EMG signals, it may provide additional information about the strategies used by subjects under well-controlled static conditions.

The MultiLEIT consists of a rigid aluminum structure instrumented with two 6-DOF force/torque sensors positioned above the knee and below the ankle, as shown in Figures 1 and 2.

The 12-DOF measurements are converted through Jacobian transformations to the eight torques in the hip (3), knee (3) and ankle (2). The MultiLEIT can measure maximal and submaximal torque generation in either leg of individuals from around 150 cm (for reference, the 5th percentile in height for women is 153.8cm) to 190 cm (just above the 95th percentile of height in men which corresponds to 187.3 cm) ³⁶. The MultiLEIT can be used in the upright position but also in various angles of inclination and in a supine position.

A. Design Description

a) Adjustability—The device can be adjusted to fit subjects up to the 95th percentile weight in males (130.5 kg) and the 95th percentile waist circumference (133.9 cm) ³⁶. Subjects are seated on a gel-padded bicycle seat that provides partial gluteal support while the pelvis remains in a neutral tilt. A harness with adjustable straps is then placed around the trunk to secure subjects into the setup. Padded-clamps are then positioned over each shoulder, to keep subjects from pushing in the upward direction during maximum voluntary torque (MVT). Pelvic clamps secure the pelvis and prevent subjects from pivoting on the bicycle seat. Once the subject's upper body has been securely fit into the setup, the tested lower extremity is positioned onto the instrumented portion of the MultiLEIT. The system is adjusted to bring the 6-DOF sensors to the desired location. Adjustability of sensor location in the anterior-posterior (x), superior-inferior (y) and medial-lateral (z) directions is achieved using a sliding clamp system that constitutes the device's main structure (Figure 2). The top sensor is attached to a rigid cuff padded with memory foam. The thigh is secured inside this cuff until there is no slack between the thigh and the foam. This attachment point was selected to maximize rigid coupling with the setup despite changes in muscle girth during contraction of the quadriceps and hamstring muscles (any slack in the thigh-cuff interface can introduce small changes in the joint angles during the experiment resulting in measurement errors). The foot is placed onto the rigid aluminum foot-plate that matches the subject's foot size. The foot is then coupled to the setup using rigid fiber-glass cast (Össur, Orange County, CA) around the foot and the foot-plate, which is then bolted onto the MultiLEIT.

b) Measurement of forces and torques—The device was designed to quantify static joint torque coupling patterns at the hip, knee and ankle in 8-DOF (hip abduction/adduction, hip flexion/extension, hip external/internal rotation, knee flexion/extension, knee valgus/varus-abduction/adduction-, knee internal/external rotation, ankle dorsi/plantar flexion). The top sensor is positioned laterally above the knee with its face parallel to the sagittal plane. Measurements from this sensor will be used to calculate torques generated at the hip joint. The bottom sensor is located on a parallel plane to the top sensor, on the lateral edge of the foot. The bottom sensor is used to measure torques generated at the ankle and knee joints and a portion of the hip torques.

The 6-DOF force/torque sensors have a ± 2000 N force range in F_x and F_y and ± 4000 N F_z range and a torque range of ± 400 Nm in x , y and z (Model 75E20A4, JR3 Inc., Woodland, CA) to accommodate the 95th percentile maximum extension torque that can be generated at the hip by males (419 Nm - ³⁷). Reported sensor resolution is 0.25N in F_x and F_y , 0.5 N in F_z and 0.05Nm in T_x , T_y and T_z . Error introduced in the joint torque

calculations due to sensor resolution is negligible. Raw forces and torques measured from the sensors are transformed to joint torques and displayed on a monitor to provide visual feedback for the subject. The free body diagram for the system is shown in figure 2. Overall dimensions of the system are 2150mm×1605mm×1200mm (height × depth × width).

c) Decoupling of tested lower extremity—The design of the MultiLEIT is intended to mechanically and neurally decouple the tested lower extremity from the body so that the measured joint torques only reflect the joint torques generated with the tested leg. Also, the subject will perceive the non-tested leg as if it was in swing phase of gait. The upper body attachment system limits trunk and pelvis motion that can affect the measurements. Neuro-mechanical decoupling of the tested leg from the non-tested leg is achieved by rigidly attaching the tested lower extremity to the sensors while the contralateral leg is left floating, placed on an elastic sling. This eliminates torque contributions of the non-tested leg to the measurements and generation of involuntary torques in the tested leg due to inter-limb coupling driven by the non-tested leg³⁸. The length of the sling is adjusted to the length of the subject's leg so that the leg is fully extended; this prevents subjects from pushing against the support surface, i.e, removing the mechanical ground for the contralateral lower extremity.

d) Comfort and safety—The MultiLEIT is instrumented with an electrical actuation mechanism that can recline the subject to a horizontal position, removing body weight support from the bicycle seat and allowing the subject to rest. Accessibility to the setup for severely impaired subjects is achieved using a stepping system that can be adjusted in height to motor capabilities of each subject. Safety is guaranteed by allowing subjects to exit the setup in in less than one minute in case of an emergency, using quick release mechanisms at all attachment points.

B. Computation of Joint Torques

Based on the assumption of a static system, the principle of virtual work for static equilibrium³⁹ was used to calculate the joint torques as a function of the forces and torques measured at each load cell:

$$\boldsymbol{\tau} = \mathbf{J}^T(\mathbf{q}) [\mathbf{F}^T \mathbf{T}^T]. \quad (\text{Equation 1})$$

$\mathbf{J}^T$ is the Jacobian transformation matrix determined by the geometry of the system. If $\mathbf{F}$ and $\mathbf{T}$ are the x , y and z forces and torques measured by the sensor, joint forces and torques are defined as:

$$\begin{bmatrix} F_x \\ F_y \\ F_z \\ T_x \\ T_y \\ T_z \end{bmatrix} = \begin{bmatrix} \mathbf{R}(\theta) & \mathbf{zeros}(3) \\ \begin{bmatrix} 0 & -d_z & d_y \\ d_z & 0 & d_x \\ -d_y & -d_x & 0 \end{bmatrix} & \mathbf{R}(\theta) \end{bmatrix} * \begin{bmatrix} F_{xJR3} \\ F_{yJR3} \\ F_{zJR3} \\ T_{xJR3} \\ T_{yJR3} \\ T_{zJR3} \end{bmatrix} \quad \text{Equation 2}$$

where, $\mathbf{R}(\theta)$ is the 3×3 rotation matrix defining the orientation of the sensor coordinate system with respect to the joint coordinate systems and d_i , for $i=x, y$ or z is the distance between the geometrical origin of the sensor and the origin of each joint center of rotation in the corresponding direction. The angles for the rotation matrices are defined in Figure 2 and Table 1.

The knee and ankle **net** joint torques only include forces measured at the **bottom** load cell, already transformed to the center of rotation of each joint. Definition of each joint's coordinate system is presented in Table 1. Joint forces and torques for the ankle (**A**) and knee (**K**) are defined in equations 3 and 4 respectively. $\mathbf{R}_x(\theta_{HABD})$ indicates a rotation around the x-axis for the corresponding hip abduction angle. d_x , d_y and d_z are the distances from the center of the bottom JR3 to the center of rotation of the ankle. $\mathbf{R}_z(\theta_a)$ is the rotation matrix around the z axis of a magnitude equal to the ankle plantarflexion angle. $\mathbf{R}_z(\theta_k)$ is the rotation around z of a magnitude corresponding to the knee flexion angle. l_s is the shank length from the ankle center of rotation to the knee center of rotation. These transformations are applied to the forces and torques measured by the bottom sensor.

The net hip torque (**H**) is computed using homogeneous transformations from both load cells to the hip joint's center of rotation. Unavoidable small movements of the leg occur in the system when there is a change in muscle girth after contraction (see Results). When this occurs, the bottom load cell will measure a portion of the reaction torques in response to torques generated at the hip. In this case, the contribution of the bottom load cell to the total hip torque will differ from zero. If the system is completely static (minuscule changes in muscle girth), the hip forces and torques measured in the bottom load cell and transformed to the hip will be zero.

In Equation 5, d_x , d_y and d_z are the distances from the center of the top JR3 to the center of rotation of the knee. In the case of a completely static system with no changes in muscle girth during MVT, the matrix of knee forces and torques (**K**) transformed to the hip coordinate system will be close to zero. $\mathbf{R}_z(\theta_h)$ is the rotation around z of a magnitude corresponding to the hip flexion angle. l_t is the thigh length from the knee center of rotation to the hip center of rotation. $\mathbf{F}_T^T$ and $\mathbf{T}_T^T$ are the transposed forces and torques measured by the top JR3 in x, y and z.

Note that the gravitational force due to the mass of the leg measured by the sensors was not included in the equations. Before each trial, subjects were asked to relax, and the forces and torques recorded on the sensors during this period were used to baseline correct the data.

$$\mathbf{A} = \begin{bmatrix} \mathbf{R}_x(\theta_{HABD}) & \mathbf{zeros}(3) \\ \begin{bmatrix} 0 & -d_z & d_y \\ d_z & 0 & d_x \\ -d_y & -d_x & 0 \end{bmatrix} & \mathbf{R}_x(\theta_{HABD}) \end{bmatrix} * \begin{bmatrix} \mathbf{R}_z(\theta_a) & \mathbf{zeros}(3) \\ \mathbf{zeros}(3) & \mathbf{R}_z(\theta_a) \end{bmatrix} * \begin{bmatrix} \mathbf{F}_B^T & \mathbf{T}_B^T \end{bmatrix} \quad (\text{Equation } 3)$$

$$\mathbf{K} = \begin{bmatrix} \mathbf{R}_x(\theta_{HABD}) & \mathbf{zeros}(3) \\ \begin{bmatrix} 0 & -d_z & d_y \\ d_z & 0 & d_x \\ -d_y & -d_x & 0 \end{bmatrix} & \mathbf{R}_x(\theta_{HABD}) \end{bmatrix} * \begin{bmatrix} \mathbf{R}_z(\theta_k) & \mathbf{zeros}(3) \\ \begin{bmatrix} 0 & 0 & l_s \\ 0 & 0 & 0 \\ l_s & 0 & 0 \end{bmatrix} & \mathbf{R}_z(\theta_k) \end{bmatrix} * [\mathbf{F}_B^T \quad \mathbf{T}_B^T] \quad (\text{Equation } 4)$$

$$\begin{aligned} \mathbf{H} = & \begin{bmatrix} \mathbf{R}_z(-\theta_k) & \mathbf{zeros}(3) \\ \begin{bmatrix} 0 & 0 & l_t \\ 0 & 0 & 0 \\ -l_t & 0 & 0 \end{bmatrix} & \mathbf{R}_z(-\theta_k) \end{bmatrix} * \mathbf{K}^T \\ & + \begin{bmatrix} \mathbf{R}_x(\theta_{HABD}) & \mathbf{zeros}(3) \\ \begin{bmatrix} 0 & -d_z & d_y \\ d_z & 0 & d_x \\ -d_y & -d_x & 0 \end{bmatrix} & \mathbf{R}_x(\theta_{HABD}) \end{bmatrix} * \begin{bmatrix} \mathbf{R}_z(\theta_h) & \mathbf{zeros}(3) \\ \begin{bmatrix} 0 & 0 & l_t \\ 0 & 0 & 0 \\ -l_t & 0 & 0 \end{bmatrix} & \mathbf{R}_z(\theta_h) \end{bmatrix} * [\mathbf{F}_T^T \quad \mathbf{T}_T^T] \end{aligned} \quad (\text{Eq } 5)$$

III. Methods

A. Device Validation

Validation of the setup was performed using a 4-link (trunk, thigh, leg, foot) joint torque mechanic validation system made of aluminum beams connected through two hinge joints to simulate the ankle and knee and a rotatory joint for the hip. The aluminum structure was attached to the setup at both sensors using bolts above the knee hinge joint and at the foot segment of the aluminum leg to mimic a typical subject's configuration. Muscles crossing the hip and knee joint were simulated using springs with known elasticity constants and known moment arms to the center of rotation of each joint for flexion torques. A one-DOF force sensor was used to apply a known force at a known lever arm from the rotatory hip joint to generate a hip adduction torque. The magnitude of the torques generated at each joint could therefore be calculated. Torques were generated at the hip, knee or ankle joint and then simultaneously at all joints by engaging the springs first separately and then simultaneously. Five trials were acquired for each task. Sensor forces and torques were transformed into joint torques. Trials for each task were averaged to eliminate random noise.

B. Sensitivity analysis

Monte Carlo simulations⁴⁰ were run on the geometric transformation equations for joint-torque calculations, on both the joint torque mechanic validation data and on the experimental data on one control subject for the hip abduction MVT task. Parameters were varied independently and simultaneously in the simulations, including the joint angles (θ_{HABD} , θ_h , θ_k , θ_a) and the distance offsets from the center each JR3 to the centers of rotation of the ankle, knee and hip in x , y and z . All sensitivity analyses were run in Matlab. The Jacobian transformation parameters were selected from a probability density function defined as:

$$q = \mu + \text{randN} \times SD, \text{ for each parameter } q.$$

μ is the measured parameter and $randN$ is a randomly generated number from a normal distribution with standard deviation (SD) set to ± 25 mm for all distances measured (segment and offset from load cells) and $\pm 10^\circ$ for joint angles. Monte Carlo simulations were run for 100,000 iterations.

C. Experimental Protocol

Testing on post-stroke individuals focused on hip abduction/adduction and flexion/extension, knee flexion/extension and ankle dorsi/plantarflexion. Coupling of hip, knee and ankle extension with hip adduction is observed clinically in individuals post-stroke and is commonly known as the “extensor synergy”³⁰. We tested the measurement capabilities of the MultiLEIT to quantify joint-torque coupling and compared our results with clinical observation³⁰ and previous research^{17–20}.

Two moderately impaired individuals post-stroke (S1, female, age=60 years, time since stroke=6 years, FMA=20, weight=190lbs, height=5ft 1in. S2, male, age=53 years, time since stroke=3 years, FMA=17, weight=185lbs, height=5ft 10in) were recruited from an Institutional Review Board-approved database, the Clinical Neuroscience Research Registry (CNRR). All experimental protocols were approved by Northwestern University’s Institutional Review Board. Subjects were examined by a licensed physical therapist to verify admissibility to the study. Initial evaluation of lower extremity motor function was done using the Fugl-Meyer Motor Assessment (FMA)⁴¹. Two healthy gender and age matched control subjects (C1, female, weight=130lbs, height=5ft 6in, age 58. C2, male, weight=145lbs, height 5ft 7in, age 52) were recruited for this study. All subjects participated in the single DOF maximal and submaximal protocol and dual DOF protocol.

Subjects were fitted into the MultiLEIT and the lower extremity was secured into the device. Subjects were asked to perform a single-task protocol consisting of maximum voluntary torques (MVT) in either hip abduction/adduction, hip flexion/extension, knee flexion/extension or ankle dorsi/plantar flexion in randomly ordered blocks to discard learning effects. Primary (voluntary and instructed) torques and secondary (concurrent and spontaneous) torques were computed online based on the raw forces and torques measured from the two 6-DOF load cells. Visual feedback of the DOF being maximized was provided as shown in Figure 3. No instructions were given for any of the other DOFs. In each trial, the subject was asked to start in a relaxed state, ramp up to maximal torque production and sustain it for 2 seconds. A trial was considered successful if the MVT plateaued for at least 250ms. Two trials with MVT values within 10% of each other, with the second trial being of smaller magnitude than the first trial were required to ensure MVTs were achieved. Only two MVT trials were required to avoid fatigue. The variables of interest in this study were the MVT magnitude for a 250ms time window in the instructed DOF and the torques at the other joints and DOFs in the same time window (Figure 4).

For the submaximal and the dual-task protocols, subjects were asked to generate submaximal hip extension torques (25, 50, 75% MVT) and then in separate trials to combine them with 50% of their hip abduction MVT. Visual feedback consisted of a moving speedometer with the vertical displacement mapped to hip extension and the horizontal displacement mapped to the hip abduction/adduction torque (Figure 3).

D. Data processing and analysis

Data acquisition was performed using a National Instruments Single Differential Legacy Device (PCI 6031E, National Instruments, Austin, Texas). Digital signal processing and data transformation was performed in a custom made Matlab graphical user interface (GUI) (Mathworks Inc., Natick, MA). Data were acquired for 8s using a 1000Hz sampling frequency. Force and torque data were filtered using a 250ms moving average window before transforming it to each joint. All joint torques at the time point of MVT in the primary direction were extracted and normalized to the global maximum torque obtained across all trials (Figure 4). The maximum (voluntary or spontaneous) torque for each degree of freedom was extracted from the entire session and all torques were expressed as a percentage of the global maximum.

E. Effect of muscle girth changes on limb movement and torque estimation

We quantified the change in parameters resulting from limb movement due to changes in soft tissue girth during muscle contraction using a motion analysis system (Optotrak Certus, Northern Digital Incorporated). We placed three infrared markers on the tested lower extremity: marker 1 was located on the center of the patella (corresponding to the frontal plane origin of the center of rotation of the knee). Marker 2 was placed on the medial surface of the knee (sagittal plane origin of the knee center of rotation). Marker 3 was located on the midpoint of the line connecting the hip and knee joints on the surface of the thigh. The location of the markers was recorded simultaneously with the MVT data.

IV. Results

A. Validation Results

1) Quantification of known torques—For the validation procedure using the mechanic validation system the torques generated at each joint were 35 Nm at the hip for the flexion DOF, 36Nm at the hip for adduction, 17.5 Nm at the knee for flexion and 15 at the ankle for dorsiflexion. The variance of the output (VAF – calculated torques) accounted for almost 100% of the input (applied torques – VAF = 99.92%). The measurement error was calculated for each joint as the difference between the input and the measured torque. For the hip abduction/adduction, the error was 1.717%, for hip flexion/extension it was calculated to be 1.9%, for knee flexion/extension it was 2.5% and for the ankle it was 0.78%.

2) Changes in joint angle—Participants are attached to the setup at the trunk and at two points on the lower extremity, so that no visible movement of the segments can be detected when the muscles are relaxed. However, changes in muscle girth are inevitable when subjects generate maximum hip and knee extension torques. Particularly in healthy control individuals, these changes can be substantial. Measurements obtained from the motion analysis protocol showed the greatest displacement for marker 1 (center of patella) during knee extension (36 mm displacement in the x direction - posteriorly) and hip extension (40 mm displacement in the x direction - posteriorly). Changes in muscle girth generated changes in hip flexion angle of less than 4° and changes in knee angle of less than 5°.

The above changes in torque angle were used simultaneously in the mechanic leg validation system to identify the changes in the torque generated, given a known torque. Offsets for the measured angle due to changes in muscle girth were accounted for errors in measurement of the hip abduction/adduction torque of 5%, of the hip flexion/extension torque of 4%, attenuation of the knee flexion torque by 12% and no error in the ankle torque.

B. Sensitivity Analysis Results

1) Sensitivity analysis for the mechanic leg simulation—For the mechanic leg validation, variation from real joint segment lengths of ± 25 mm generated errors in torque measurement of up to 4% for the knee flexion/extension torque, 3.5% for the hip flexion/extension torque and up to 2.9% for the hip adduction torque. Results for variations in the distance offsets from the bottom JR3 sensor generated substantial errors in torque measurement. Measurement error in the vertical distance (y) between the bottom JR3's origin and the center of rotation of the ankle caused errors in knee torques close to 18%. In contrast, errors in offset for the top JR3 are less critical; offsets of ± 25 mm from the real distances generate errors of $\pm 2\%$ in hip torques.

On the other hand, errors in measurement of joint angles of $\pm 10\%$ had a substantial effect on knee flexion/extension torques (up to 42% for an error in measurement of the hip flexion angle). As an end result, these errors can alter the directionality of the calculated torque for small torque values. Errors in measurement of the remaining angles and offsets generated errors smaller than 2%.

C. Experimental Results

Figure 5 shows the results from the single-task protocol for the paretic and non-paretic extremities of S1 and S2 and for the right extremity of C1 and C2. The bars represent the normalized torque produced for each task in each degree of freedom (series labels). Preliminary results for the hip extension MVT task (top panel) show hip extension MVTs spontaneously coupled with hip adduction torques of over 75%MVT and with extension of all other joints on the paretic and non-paretic extremities of both individuals post-stroke.

Hip extension/adduction coupling was also observed during the ankle plantarflexion task for both individuals post-stroke. On both paretic extremities, maximum ankle plantarflexion torques were obtained as a spontaneous torque during hip extension, i.e. subjects were unable to volitionally generate the maximum ankle plantarflexion torque during this task, but could generate maximum plantarflexion as a secondary torque during hip extension.

Coupling patterns of C1 and C2 differed from those seen in the paretic extremity (Figure 5). C1 coupled extension with abduction and C2 coupled adduction/extension with dorsiflexion. For the ankle plantarflexion MVT task, C1, C2 and the non-paretic extremity of S2 coupled ankle plantarflexion with hip abduction, hip extension and knee flexion.

For the submaximal protocol, results indicated extension/adduction coupling, proportional to the percentage of hip extension in both S1 and S2. For control individuals, C1 coupled hip extension with hip abduction at submaximal levels and C2 generated small coupling at other degrees of freedom during submaximal hip extension.

During the implementation of the dual-task protocol, C1 and C2 were able to generate the instructed hip abduction torque, with no influence of the percentage of hip extension generated. In contrast, S1 and S2 were unable to combine the requested torques (Figure 6). As the level of hip extension increased, spontaneous coupling with hip adduction occurred, decreasing the net hip abduction torque participants were able to generate. For levels of hip extension above 25%, the spontaneous coupling overrides the volitional hip abduction torque.

Results demonstrate the ability of the MultiLEIT to measure lower extremity joint torques during single- or dual-task protocols at maximal or submaximal torque levels under well-controlled conditions.

1) Sensitivity analysis on experimental data—The sensitivity analysis on experimental data of the control subject's hip abduction MVT task, showed no effects on torque directionality. The measured MVT was 112.25Nm. For an error in measurement of the hip abduction angle of $\pm 10^\circ$, the hip abduction torque varied from 68 to 155Nm, which corresponds to an error of $\pm 39\%$ of the calculated torque. The ankle torque varied ± 9 Nm from a real value of 24Nm, corresponding to an error of $\pm 37\%$. For C1, an error in measurement of the hip abduction angle of this magnitude does not affect the directionality of the hip torque, it only attenuates or augments the torque magnitude. As mentioned previously for the joint torque mechanic leg validation, for smaller hip torques these changes in torque directionality are more likely to occur for 10° changes in joint angle (Figure 7, A). Realistically, an error in measurement of the hip abduction angle of 10° is unlikely if the measurements are performed carefully⁴². Anatomical landmarks to measure hip flexion/extension angles are easily identifiable. In order to minimize measurement error, two researchers performing the experiment will verify all measurements.

Finally, results also indicate that both the MultiLEIT and the tested lower extremity can be assumed to be in static equilibrium, since the small movements generated due to changes in muscle girth do not generate changes in joint angles greater than 5° , which account for errors in torque measurement of up to 12%.

V. Discussion and Conclusions

In this manuscript we introduce the MultiLEIT, an isometric device capable of quantifying lower extremity joint torque coupling patterns in all available degrees of freedom from hip to ankle during volitional generation of maximal and submaximal torques about a given joint. Information obtained from studies employing the MultiLEIT can help understand the neuromuscular strategies implemented at the lower extremity by a study participant and how they can differ when the neural and/or the muscular component is affected due to pathology.

Validation of the MultiLEIT was performed using both known inputs to the system and experimental data. Results for the validation procedure with known inputs demonstrate the MultiLEIT's ability to accurately measure torques generated simultaneously at multiple joints using static Jacobian transformations from two 6DOF load cells to each joint. Sensitivity analysis on these data showed that large measurement errors of hip abduction and

hip flexion angles can result in flipping of the direction of the calculated torques (for example, a positive flexion torque may be calculated when the actual input torque was a negative extension torque) particularly for torques of small magnitude. However, errors of this magnitude ($\pm 10^\circ$) are highly unlikely given that the goniometric measurements are performed carefully by experienced experimenters⁴². Some of these issues could potentially be avoided if an ideal system was built with rigid materials and instrumented with sensors aligned with each joint's center of rotation and with the joint rotational axes to quantify the torques generated at each joint. In practice however, the inherent area of interface between the leg and the attachment, the rigidity of materials needed and the potential contact with bony prominences makes this ideal system impossible to build. The MultiLEIT system addresses this by including six degree of freedom sensors to quantify the reaction forces generated at the sensors during maximal and submaximal torque generation at each joint and allow measurement of individual joint torques in a highly coupled system.

Motion analysis during the experiment was run on C1 to test sources of potential movement of the leg/MultiLEIT during the experiment. We quantified marker movement due to changes in muscle girth during maximum muscle contraction. Particularly, the patellar marker displacement was quantified to be less than 40mm, which generated changes in joint angles of less than 5° for the hip and knee flexion angles. These values can be considered negligible and they do not affect the calculated joint torques. Therefore, the system can be assumed to be a static system for analysis purposes.

Experimental data was collected on individuals of different sizes (C1: 130lbs, 5ft6in, C2: 145lbs, 5ft7in and S1: 190lbs, 5ft 1in, S2: 185lbs, 5ft10in) for both right and left lower extremities. The MultiLEIT was successfully adjusted to fit both individuals. The torques measured at each joint were in accordance to those reported in the literature and previous research^{17–20, 37}. Results for C1 and C2 quantified joint torque coupling patterns consistent with the mechanical coupling of the lower extremity due to biarticular muscles¹⁴ and learned motor behaviors^{43,44}. Results obtained for S1 and S2 indicated the expression of joint torque coupling patterns identical to those described clinically in the extensor synergy³⁰. These stereotypical coupling was observed for maximal and submaximal levels of hip extension and for maximal ankle plantarflexion.

Future research using the MultiLEIT will continue to quantify changes in joint torque coupling patterns in post-stroke individuals with different levels of impairment. We intend to quantify subject's ability to generate coupling patterns that are outside those observed in the clinic to determine whether all individuals post-stroke, with varying impairment levels are constrained to coupling patterns described in the extension synergy. Future work will also assess abnormal joint torque coupling in children with hemiplegic cerebral palsy, in order to assess the relationship between time of the lesion, abnormal joint torque coupling and distribution of weakness in the lower extremity. Information obtained from studies using the MultiLEIT can provide evidence on the specific joint-torque weakness and abnormal coupling patterns developed after changes in supraspinal drive or as strategies used to prevent joint pain or compensate for muscle weakness. This information can aid in the design of rehabilitation approaches to address the underlying mechanisms responsible for

movement abnormalities, such as abnormal joint torque coupling that may occur in the lower extremity.

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References

1. Wu G, Siegler S, Allard P, et al. ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion--part I: ankle, hip, and spine. International Society of Biomechanics. *Journal of biomechanics*. 2002; 35:543–8. [PubMed: 11934426]
2. Brault, MW. U.S. Census Bureau. Americans with disabilities, 2005. Washington, D.C: U.S. Dept. of Commerce, Economics and Statistics Administration, U.S. Census Bureau; 2008. p. 23
3. Dyer JO, Maupas E, Melo SD, Bourbonnais D, Forget R. Abnormal coactivation of knee and ankle extensors is related to changes in heteronymous spinal pathways after stroke. *Journal of Neuroengineering and Rehabilitation*. 2011;8. [PubMed: 21291548]
4. Chow JW, Stokic DS. Force control of quadriceps muscle is bilaterally impaired in subacute stroke. *J Appl Physiol*. 2011; 111:1290–5. [PubMed: 21885803]
5. Pohl PS, Startzell JD, Duncan PW, Wallace D. Reliability of lower extremity isokinetic strength testing in adults with stroke. *Clin Rehabil*. 2000; 14:601–7. [PubMed: 11128734]
6. Lewek MD, Breslin R, Hlad L, Lanton A, John JS. Non-paretic quadriceps activity influences paretic quadriceps activity post-stroke. *Clin Neurophysiol*. 2010; 121:1962–7. [PubMed: 20451444]
7. Eng JJ, Lomaglio MJ, Macintyre DL. Muscle Torque Preservation and Physical Activity in Individuals with Stroke. *Med Sci Sports Exerc*. 2009; 41:1353–60. [PubMed: 19516167]
8. Dalgas U, Stenager E, Lund C, et al. Neural drive increases following resistance training in patients with multiple sclerosis. *J Neurol*. 2013
9. Sandroff BM, Sosnoff JJ, Motl RW. Physical fitness, walking performance, and gait in multiple sclerosis. *J Neurol Sci*. 2013; 328:70–6. [PubMed: 23522499]
10. Thompson N, Stebbins J, Seniorou M, Newham D. Muscle strength and walking ability in diplegic cerebral palsy: implications for assessment and management. *Gait Posture*. 2011; 33:321–5. [PubMed: 21169021]
11. Costa PB, Ryan ED, Herda TJ, et al. Acute effects of static stretching on peak torque and the hamstrings-to-quadriceps conventional and functional ratios. *Scand J Med Sci Sports*. 2013; 23:38–45. [PubMed: 21672027]
12. Grood ES, Suntay WJ. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *J Biomech Eng*. 1983; 105:136–44. [PubMed: 6865355]
13. Andrews AW, Bohannon RW. Distribution of muscle strength impairments following stroke. *Clin Rehabil*. 2000; 14:79–87. [PubMed: 10688348]
14. Zajac FE. Muscle coordination of movement: a perspective. *Journal of biomechanics*. 1993; 26 (Suppl 1):109–24. [PubMed: 8505346]
15. Zajac FE, Neptune RR, Kautz SA. Biomechanics and muscle coordination of human walking. Part I: introduction to concepts, power transfer, dynamics and simulations. *Gait Posture*. 2002; 16:215–32. [PubMed: 12443946]
16. Kuypers, HGJM. The descending pathways to the spinal cord, their anatomy and function. In: Eccles, JC.; Schade, JP., editors. *Organization of the spinal cord Progress in brain research*. Amsterdam: Elsevier; 1964.
17. Cruz TH, Dhaher YY. Evidence of abnormal lower-limb torque coupling after stroke: an isometric study. *Stroke*. 2008; 39:139–47. [PubMed: 18063824]
18. Hidler JM, Carroll M, Federovich EH. Strength and coordination in the paretic leg of individuals following acute stroke. *IEEE Trans Neural Syst Rehabil Eng*. 2007; 15:526–34. [PubMed: 18198710]

19. Neckel N, Pelliccio M, Nichols D, Hidler J. Quantification of functional weakness and abnormal synergy patterns in the lower limb of individuals with chronic stroke. *J Neuroeng Rehabil.* 2006; 3:17. [PubMed: 16857059]
20. Thelen DD, Riewald SA, Asakawa DS, Sanger TD, Delp SL. Abnormal coupling of knee and hip moments during maximal exertions in persons with cerebral palsy. *Muscle Nerve.* 2003; 27:486–93. [PubMed: 12661051]
21. Cruz TH, Lewek MD, Dhaher YY. Biomechanical impairments and gait adaptations post-stroke: multi-factorial associations. *J Biomech.* 2009; 42:1673–7. [PubMed: 19457488]
22. Beer, RF.; Dewald, JP.; Given, JD.; Schultz, LJ. Society for Neuroscience. 1995. Quantification of weakness and muscle synergies in hemiparetic stroke subjects: preliminary results.
23. Dewald JP, Beer RF. Abnormal joint torque patterns in the paretic upper limb of subjects with hemiparesis. *Muscle Nerve.* 2001; 24:273–83. [PubMed: 11180211]
24. Dewald JP, Sheshadri V, Dawson ML, Beer RF. Upper-limb discoordination in hemiparetic stroke: implications for neurorehabilitation. *Top Stroke Rehabil.* 2001; 8:1–12. [PubMed: 14523747]
25. Dewald JPA, Ellis MD, Acosta AM, Yao J. Position-dependent torque coupling and associated muscle activation in the hemiparetic upper extremity. *Experimental Brain Research.* 2007; 176:594–602. [PubMed: 16924488]
26. Dewald JPA, Pope PS, Given JD, Buchanan TS, Rymer WZ. Abnormal Muscle Coactivation Patterns during Isometric Torque Generation at the Elbow and Shoulder in Hemiparetic Subjects. *Brain.* 1995; 118:495–510. [PubMed: 7735890]
27. Twitchell TE. The restoration of motor function following hemiplegia in man. *Brain.* 1951; 74:443–80. [PubMed: 14895765]
28. Sukal-Moulton T, Murray TM, Dewald JPA. Loss of independent limb control in childhood hemiparesis is related to time of brain injury onset. *Experimental Brain Research.* 2013; 225:455–63. [PubMed: 23411673]
29. Reinkensmeyer DJ, Kahn LE, Averbuch M, McKenna-Cole A, Schmit BD, Rymer WZ. Understanding and treating arm movement impairment after chronic brain injury: Progress with the ARM guide. *Journal of Rehabilitation Research and Development.* 2000; 37:653–62. [PubMed: 11321001]
30. Brunnstrom, S. Movement therapy in hemiplegia: a neurophysiological approach. 1. New York: Medical Dept; 1970. p. 192
31. Brown DA, Kautz SA, Dairaghi CA. Muscle activity patterns altered during pedaling at different body orientations. *Journal of Biomechanics.* 1996; 29:1349–56. [PubMed: 8884480]
32. Davis BL, Cavanagh PR. Simulating Reduced Gravity - a Review of Biomechanical Issues Pertaining to Human Locomotion. *Aviat Space Envir Md.* 1993; 64:557–66.
33. Dietz V, Gollhofer A, Kleiber M, Trippel M. Regulation of Bipedal Stance - Dependency on Load Receptors. *Experimental Brain Research.* 1992; 89:229–31. [PubMed: 1601101]
34. Gutierrez GM, Chow JW, Tillman MD, McCoy SC, Castellano V, White LJ. Resistance training improves gait kinematics in persons with multiple sclerosis. *Arch Phys Med Rehab.* 2005; 86:1824–9.
35. Lo AC, Triche EW. Improving gait in multiple sclerosis using robot-assisted, body weight supported treadmill training. *Neurorehabilitation & Neural Repair.* 2008; 22:661–71. [PubMed: 18971381]
36. National Center for Health Statistics (U.S.), National Health Examination Survey (U.S.), National Health and Nutrition Examination Survey (U.S.) and Hispanic Health and Nutrition Examination Survey (U.S.). Anthropometric reference data for children and adults : United States, 2007–2010. Hyattsville, Md: U.S. Dept. of Health and Human Services, Centers for Disease Control and Prevention, National Center for Health Statistics; 2012.
37. Chaffin DB. Occupational biomechanics--a basis for workplace design to prevent musculoskeletal injuries. *Ergonomics.* 1987; 30:321–9. [PubMed: 3582344]
38. Kautz SA, Patten C. Interlimb influences on paretic leg function in poststroke hemiparesis. *Journal of neurophysiology.* 2005; 93:2460–73. [PubMed: 15590727]
39. Siciliano, B. Robotics : modelling, planning and control. London: Springer; 2009. p. xxivp. 632

40. Hamby DM. A Review of Techniques for Parameter Sensitivity Analysis of Environmental-Models. *Environ Monit Assess.* 1994; 32:135–54. [PubMed: 24214086]
41. Fugl-Meyer AR, Jaasko L, Leyman I, Olsson S, Steglind S. The post-stroke hemiplegic patient. 1. a method for evaluation of physical performance. *Scand J Rehabil Med.* 1975; 7:13–31. [PubMed: 1135616]
42. Gajdosik RL, Bohannon RW. Clinical measurement of range of motion. Review of goniometry emphasizing reliability and validity. *Physical Therapy.* 1987; 67:1867–72. [PubMed: 3685114]
43. Young RP, Marteniuk RG. Stereotypic muscle-torque patterns are systematically adopted during acquisition of a multi-articular kicking task. *Journal of biomechanics.* 1998; 31:809–16. [PubMed: 9802781]
44. Dietz V. Human neuronal control of automatic functional movements: interaction between central programs and afferent input. *Physiol Rev.* 1992; 72:33–69. [PubMed: 1731372]

Figure 1. Photograph of MultiLEIT experimental setup

A. (a) Rigid foot attachment and bottom sensor. (b) Adjustable foot-sole support for the non-tested leg consisting of a bungee cord attached to a placing surface. (c) Adjustable thigh cuff and bottom sensor. (d) Pelvic clamps. (e) Trunk harness (f) Shoulder clamps.

Figure 2. Schematic of the MultiLEIT

(a,b) Adjustable foot and thigh placement system. Thigh cuffs and foot-plate are not shown. (c) Gel padded bike seat. (d) Upper body positioning system. (d1) Hip clamping mechanism. (d2) Back plate. Harness not shown. (d3) Shoulder clamps. (e) JR3 6DoF sensors. (B). Angle descriptions, sagittal plane. The hip flexion angle is measured with respect to the projection of the trunk's gravity vector. The knee flexion angle is measured with respect to the thigh. The ankle angle is measured with respect to the shank. For the hip abduction angle (frontal plane), it is measured with respect to the gravity vector's direction. (C) Free body diagram of the system.

Figure 3. Visual feedback showing the speedometer and moving speedometer provided to the subject for the MVT task and the submaximal and dual DOF task

For MVT tasks, the speedometer displays the torque that is being maximized. Flexion/abduction are shown as counterclockwise rotation of the arrow. For submaximal and dual tasks, the vertical displacement displays the hip extension torque. For the dual task, the horizontal displacement displays the instructed hip abduction/adduction degree of freedom. The direction of movement on the horizontal axes is adjusted to the right/left extremity to correspond to the direction of torque generation thus making it more intuitive the study participant.

Figure 4. Calculation of MVTs at the primary and secondary DOF for a hip extension primary task

The MVT is identified as the maximum torque along the primary direction after signal processing. The time-point for that MVT was identified. The corresponding torque magnitudes in the other DOFs for that time point were obtained.

Figure 5. Results for the single DOF MVT task for two control and two individuals post-stroke Bar graphs show mean normalized torques (y-axes) vs. DOF (x-axes). Abduction and flexion torques are positive, extension and adduction torques are negative. Tasks shown are MVT in hip extension (top) and ankle plantarflexion (bottom) for the 2 control, 2 paretic and 2 non-paretic extremities. Differences in joint torque coupling were observed for the paretic and non-paretic extremities with respect to the control extremities. For the HEXT task, individuals post-stroke showed coupling patterns coincident to the “extensor synergy” in both the paretic and non-paretic extremities. Also, post-stroke individuals generated their maximum torques in other DOF as associated torques during the HEXT task. Extension adduction coupling was also observed for the distal APF task in the individual post-stroke. (ABD: abduction. ADD: adduction. FLEX: flexion. EXT: extension. DF: dorsiflexion. PF: plantarflexion. HEXT_C: Hip extension control. HEXT_P: hip extension paretic, HEXT_NP: hip extension non-paretic. APF_C: ankle plantarflexion control. APF_P: ankle plantarflexion paretic. APF_NP: ankle plantarflexion non-paretic).

Figure 6. Results for a HEXT submaximal task on S1 and C1

Panels A, B: single degree of freedom task with submaximal hip extension. no instructions were provided for the other degrees of freedom. Post-stroke spontaneous coupling of HEXT with HADD occurred for all levels of HEXT. Panels C, D: combined task. 25, 50 and 75% HEXT-MVT + 50% HABD MVT. Control subjects generated the requested HABD torque independently of the increase in %HEXT. Stroke subjects were constrained to “extensor synergy” for levels of HEXT of or above 25%.

Figure 7. Sensitivity analysis on pilot experimental data

Points represent the calculated torques. Error bars show standard deviations from the calculated values. Jacobian matrices and corresponding joint torques were calculated for $\theta = \theta_{\text{measure}} \pm 10^\circ$ and for distances $d_i = d_{i, \text{measured}} \pm 25\text{mm}$ for $i=x,y,z$. Errors in measurement of hip abduction or flexion angles induce errors in the calculated hip abduction/adduction torques or the calculated hip flexion/extension torques that could lead to a change of torque directionality if the torques generated are small.

Table 1

Definition of joint coordinate systems for the hip, knee and ankle.

Adapted from ¹ and ¹². The joint coordinate system is composed of two axes fixed to body segments (e_1 and e_3) and their mutual perpendicular ¹². For clarity, x , y and z are equivalent to e_1 , e_2 and e_3 respectively.

Joint	Origin	e_1	e_3	e_2
Hip ¹	Center of femoral head (estimated by palpating greater trochanter, anterior superior iliac spine). Defined with respect to top and bottom sensor by dx , dy , dz , l_s , l_t , θ_a , θ_k and θ_{HABD} .	Line parallel to the line connecting both anterior superior iliac spines passing through the origin. Flexion/extension.	Axis fixed to femur joining the mid point between lateral and medial femoral epicondyles and the origin, pointing cranially. Internal/external rotation.	Axis perpendicular to e_1 and e_3 , pointing forward. Abduction/adduction.
Knee ¹²	Midpoint between the lateral and medial epicondyles. Defined with respect to bottom sensor by dx , dy , dz , l_s , θ_{HABD} , and θ_a .	Axis perpendicular to the femoral sagittal plane crossing through the midpoint between lateral and medial epicondyles. Flexion/extension.	Axis joining the midpoint between both intercondyle eminences through the center of the ankle. Internal/external rotation of the tibia.	Axis perpendicular to e_1 and e_3 , pointing forward. Abduction/adduction.
Ankle ¹	Mid-point between the medial and lateral malleoli. Defined with respect to bottom sensor by dx , dy , dz and θ_{HABD} .	Line connecting the lateral and medial malleoli, passing through the origin. Flexion/extension.	Long axes of the tibia-fibula complex in neutral position, pointing cranially. Internal/external rotation.	Axis perpendicular to e_1 and e_3 , pointing forward. Abduction/adduction.