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# Inverse-Dynamics Based Assessment of Gait using a Robotic Orthosis

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*Abstract*- **Body-weight supported treadmill training following neurological disorders such as stroke and spinal cord injuries (SCI) has become an integral part of rehabilitation for treating gait disorders. Unfortunately techniques for selecting important training parameters, such as walking speed and body-weight support, have not been established. Here we present a 3-D inverse-dynamics based approach for evaluating an individual's ability to ambulate, in terms of evaluating the magnitude and timing of joint moments at the ankle, knee and hip. This technique, which utilizes an instrumented gait robot, allows clinicians and researchers the ability to determine the training parameters in which subjects generate joint moments at the proper phases of the gait cycle which when combined with electromyographic recordings, can help establish and then progress training parameters for individuals on a subject-by-subject basis. We believe that training subjects at their preferred walking speed and body-weight support will lead to higher functional outcomes.**

***Keywords***- **robotics, gait, stroke, spinal cord injury, assessment**

## 1. INTRODUCTION

One of the primary limitations with existing locomotor training interventions in individuals with neurological injuries is the inability to determine the training conditions in which subject's generate muscle activation patterns and joint moments appropriate to the various phases of the gait cycle. For example at the end of stance, it is critical to flex the hip and subsequently flex the knee in order to propel the leg forward. Similarly during stance, adequate hip and knee extension is necessary to maintain an upright posture. Training subjects under conditions that promote these behaviors my lead to higher levels of functional recovery.

There have been few studies attempting to investigate the influence of training conditions on locomotor ability in stroke or SCI. Lamontagne and Fung (2004) looked at locomotor patterns in stroke patients at varying speeds and found that as walking speed increased, subjects demonstrated improvements in symmetry, muscle activity and kinematic patterns. Visintin and Barbeau (1994) looked at the influence of body-weight support on lower motor patterns in individuals with SCI and found that loading had strong correlations with the magnitude of muscle activity in the distal lower extremity muscles. What these and other studies have demonstrated is that altering training conditions can have a significant impact on sensory input and motor output in individuals with gait impairments due to neurological injuries such as stroke and SCI. Unfortunately these studies have primarily relied on looking at muscle activity or leg kinematics rather than joint moments. The latter is important because joint moments are directly related to powers, metabolic consumption, and efficiency (Perry, 1992).

The goal of this study was to develop and test a novel technique that allows for the computation of 3-D joint moments in individuals with severe gait impairments. This technique will allow researchers and clinicians the ability to accurately study the influence of training conditions such as speed and body-weight support, on the development of joint moments throughout the gait cycle. These measures can be combined with muscle activation patterns to determine the conditions in which an individual generates the optimal muscle activity and joint moments for any training condition, which may lead to higher gains in walking ability following locomotor training.

## 2. METHODS

### *2.1 Instrumentation*

The technique presented in this paper utilizes the Lokomat® robotic gait orthosis (Colombo et al., 2000). This system, manufactured and distributed by Hocoma AG

**Figure 1**. Lokomat robotic-orthosis (Hocoma AG, Volketswil, Switzerland)

(Volketswil, Switzerland) is comprised of a treadmill, a body-weight support system, and two light-weight robotic actuators that attach to the subject's legs (Figure 1). The Lokomat is fully programmable, including control of knee and hip range of motion, the amount of assistance the system provides to the patient, and the speed at which the patient ambulates. This high-level dynamic control is achieved by small DC motors and linear ball screw assemblies at the hip and knee joints, tightly controlled with the timing of the treadmill. The knee and hip joints have position sensors and force sensors that are monitored by the control computer throughout the training. The entire Lokomat assembly resides on a parallelogram structure which in turn is counter-balanced by a large spring. The pretension in the spring is adjusted so that the weight of the Lokomat is compensated for, preventing upward or downward forces to the subject during training.

In order to measure the ground reaction forces the subject produces while walking on the treadmill an ADAL3D-COP split-belt treadmill is used (Tecmachine, Andrezieux Boutheon, France). This treadmill contains independent belts under each foot with multi-axial force plates under each belt. The speed of each belt is controlled externally so that the belt speeds can be timed precisely with the Lokomat legs.

The standard leg attachment cuffs that come with the Lokomat were modified to contain 6-degrees of freedom load cells (JR3 Inc., Woodland, CA; Figure 2). Using the output from the load cells, interaction forces and torques generated between the subject and the Lokomat both inline with and off-axis to the intended movement can be calculated. The forces and torques recorded at each load cell are transformed back to the attachment cuff-leg interface using a homogeneous transformation (Craig, 1989).

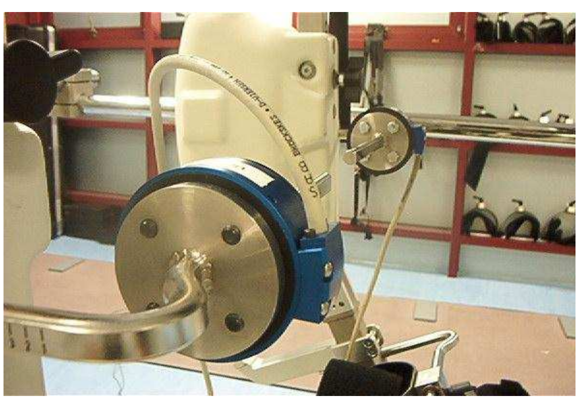

**Figure 2**. Lokomat leg cuffs shown with 6-degrees of freedom load cells

Leg kinematics are tracked with a Codomotion active marker system (Charnwood Dynamics, UK). Custom marker clusters were developed that are easily interfaced with the Lokomat leg cuffs making it easy to done and doff during setup (Figure 3).

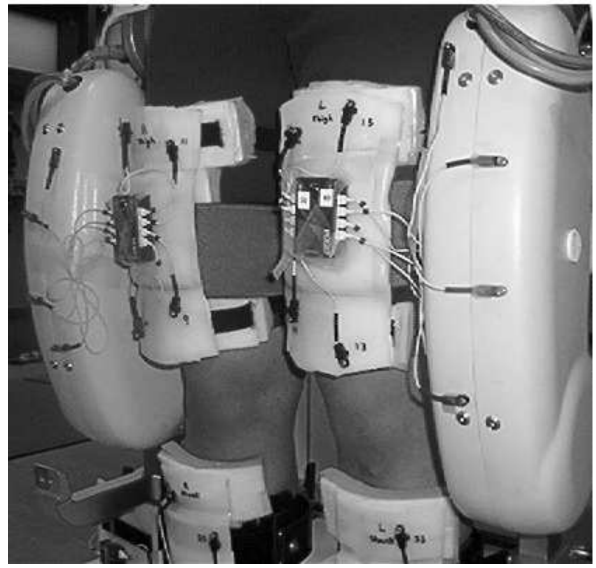

**Figure 3**. Custom marker clusters were developed for the Lokoma which rigidly couple with the leg cuffs, helping to minimize movement artifacts.

*2.2 Algorithm*

An inverse-dynamics approach is used to back calculate the joint torques developed by the subject throughout the gait cycle (Winter, 1990). Codomation marker data is first imported into Visual 3D (C-Motion, Rockville MD) where subject specific models of the lower extremities are created. The kinematics of the markers, derived from the Visual 3D model, are then exported into Matlab and combined with force plate and leg cuff load cell data to compute joint moments. Estimates of segment masses, center of masses, and moments of inertia are based on anatomical measures using the procedure described by Winter, (1990).

*2.3 Protocol*

The procedures described above were tested on a small group of healthy control subjects (n = 3). Here, each subject walked in the Lokomat at three different speeds, while at each speed, leg kinematics, ground reaction forces, and Lokomat interaction forces were collected. With each change in walking speed, the subject was allowed an acclimation phase after which 30 seconds of data were collected. No body-weight support was provided in this initial testing phase.

## 3. RESULTS

An example of a single subject's ankle, knee and hip moments generated in the sagital plane over 30 seconds of Lokomat walking are shown in Figure 4. Here we can see that the patterns are similar to those reported during over-ground gait (Perry, 1992) for ach joint. This also demonstrates that the joint moments are consistent over the multi-step sequence. The light lines illustrate individual steps while the thick line is the mean moment for each joint.

Similar patterns were demonstrated in the frontal plane. Here it is shown that joint moments are consistent and similar to those expected during over-ground gait.

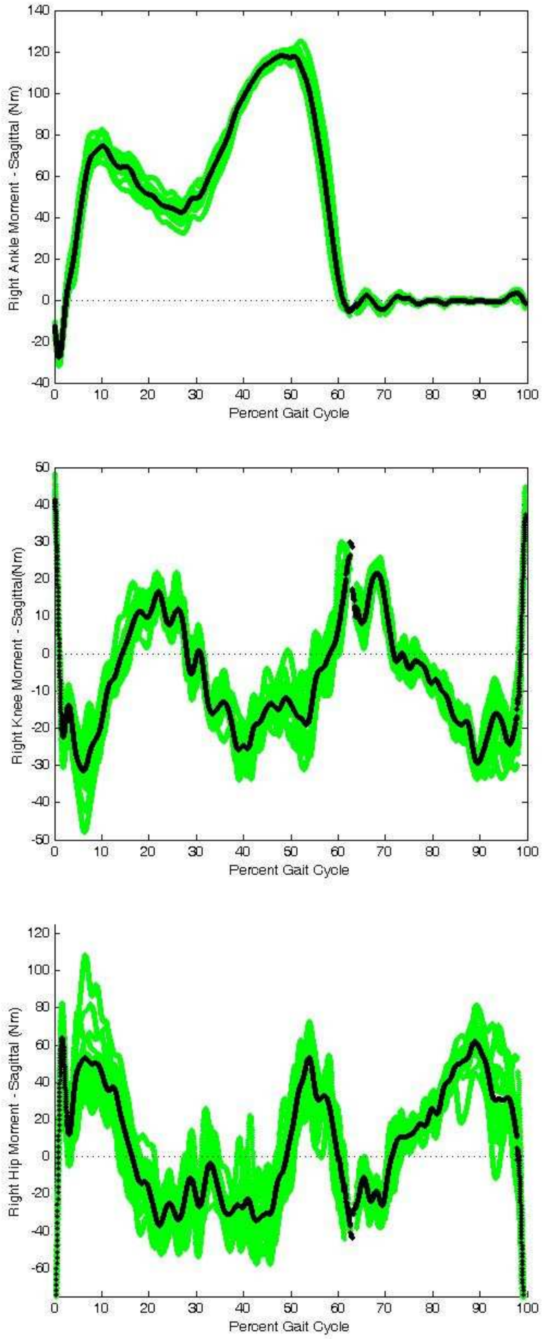


**Figure 4**. Joint moments generated in the sagittal plane at the ankle, knee and hip joints for 30-seconds of walking, along with the mean moments. Note: pos = extension.

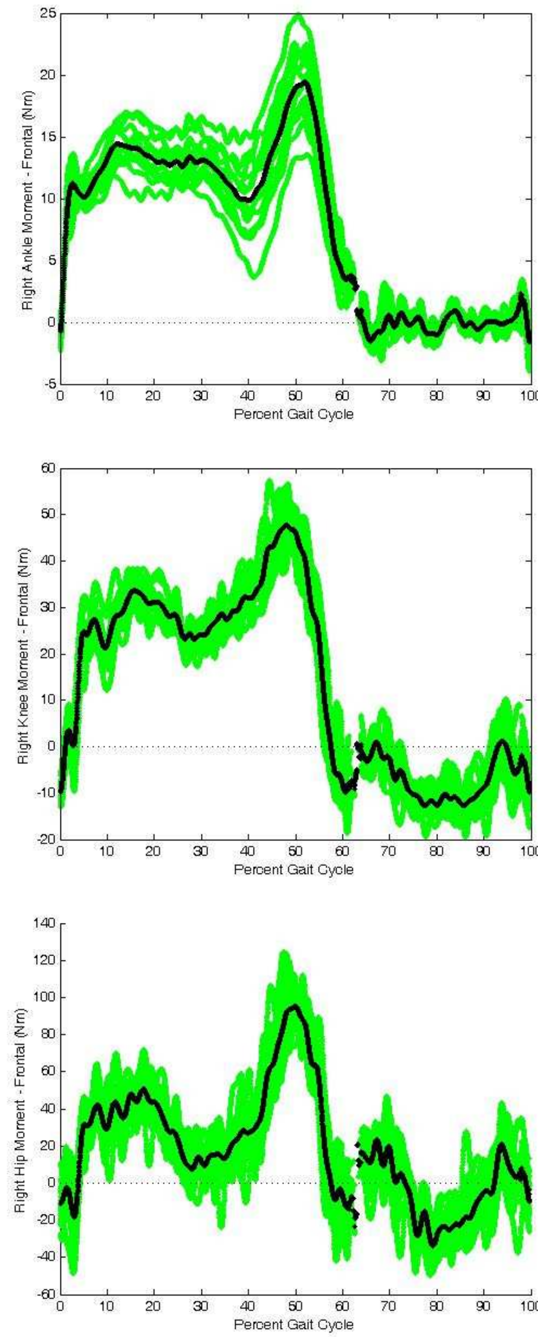


**Figure 5**. Joint moments in the frontal plane for the ankle, knee, and hip. Light traces are individual strides while the thick dark line represents the mean moment. Note: pos = abduction.

When comparisons of joint moments were made across walking speeds, it was found that joint moments scaled with the same characteristic trends as previously reported (Perry, 1992). For example, with increasing walking speed, ankle joint moments increased by 20% or more as shown in Figure 6. While the data presented here is from a healthy individual with no known gait disorders, similar behaviors are expected to be observed in individuals with gait impairments.

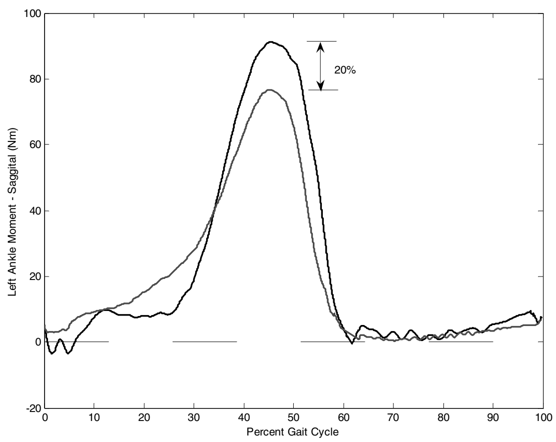


**Figure 6**. Example of increasing ankle joint moment in the sagittal plane for one subject when speed increased from 1.5 km/hr to 3.0 km/hr.

## 4. DISCUSSION

The proposed technique is now being tested at the National Rehabilitation Hospital in the Center for Applied Biomechanics and Rehabilitation Research (CABRR) in individuals with both stroke and spinal cord injuries. There, we are exploring the influence of walking speed and levels of body-weight support on enhancing joint moments during gait training. We are also investigating the presence of abnormal dynamic synergy patterns in the lower extremity of stroke patients, a behavior which has been characterized in the arms of chronic stroke subjects (Beer et al, 2000).

As outlined by previous studies, the ultimate accuracy of this technique is limited by estimates of limb segment centers of mass, inertial components, and perhaps most importantly axes of rotation. We are currently investigating the use of numerical optimization techniques to address these issues, utilizing a system identification technique based on the Lokomat instrumentation.

The algorithm can be computed in pseudo real-time so that following adjustments to the subjects training conditions, the clinician can immediately see how well the subject's step patterns adapt. This is important clinically as the number of training variables is extensive and understanding how each changes the person's ability to walk may change the outcomes of the therapeutic intervention.

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