

Effects on Normal Gait of a New Active Knee Orthosis for Hemiparetic Gait Retraining

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Abstract – Functional recovery of an impaired gait pattern is a common goal for stroke patients in their rehabilitation. Robotic and mechatronic devices offer a means of facilitating and enhancing gait retraining practices undertaken by clinicians. A new active knee orthosis has been developed for gait retraining of stroke patients that may fulfill this role. Since this device is newly developed, it is important to determine its impact on the walking patterns of healthy individuals before exploring its use in gait retraining of stroke patients. The aim of this study was to analyze adaptations in gait mechanics of healthy subjects due to the added mass of the knee orthosis while worn uni-laterally and bi-laterally. In our preliminary tests we observed significant deviations from normal gait patterns when the knee orthosis was worn uni-laterally. Conversely, minor gait deviations were seen when the knee orthosis was worn bi-laterally. This suggests that a bilateral configuration may be more suited for gait retraining purposes.

I. INTRODUCTION

Stroke is a leading cause of permanent disability in the United States. According to the National Stroke Association, each year about 730,000 people suffer a stroke, and approximately two-thirds of these individuals survive and require rehabilitation [1-3]. Approximately 80% of stroke survivors present an early motor deficit, with 50% having chronic deficits. Mobility limiting conditions such as spasticity, muscle weakness, loss of range of motion, and impaired force generation create deficits in motor control that affect the stroke survivor's capacity for independent living.

Many ambulatory stroke survivors have substantial alterations of their gait patterns as a result of hemiparesis. Compromised motor control and force generation frequently lead to limited knee flexion and stiff-legged

gait characterized by limited knee flexion during swing and typically associated with limited hip flexion and limited or absent ankle dorsiflexion.

Many rehabilitation interventions have been used to promote functional recovery in hemiparetic gait due to stroke [4,5]. Robotic and mechatronic technologies that can be integrated into portable devices and can be used by patients in the home setting are particularly attractive because they have the potential of providing tools to facilitate functional recovery, reducing cost of treatment and providing patients with adequate level of independence.

For many patients, a programmable actuated knee orthosis could guide and facilitate the recovery of a more efficient and clinically desirable gait pattern via retraining sessions. Current clinical practice is generally restricted to brief periods of less than 1 hour of gait training provided a few times per week. In between these sessions, patients continue to walk using their typical gait pattern, and likely reinforce compensatory gait patterns. Lower-extremity robotic devices for gait retraining (e.g. Lokomat®) have been developed to provide the opportunity for intense rehabilitation, but their use is limited to the clinical setting for relatively brief training sessions. A wearable training orthosis could be used by patients to guide them through a targeted gait pattern while undertaking daily activities. This strategy of reinforced therapy in a real-world environment has the potential to provide more effective gait retraining, improving one's ability to ambulate.

A new active knee orthosis, AKROD (Active Knee Rehabilitation Orthotic Device), has been designed that would provide characteristics allowing a significant improvement over existing orthotic interventions. However, before initiating testing on stroke patients, we need to assess the effect of the physical characteristics of the orthosis (e.g. alignment and mass distribution) on the gait mechanics of healthy individuals. We hypothesize that when healthy subjects wear the orthosis on one leg they will demonstrate kinetic asymmetries aimed to maintain fairly symmetric gait kinematics of. We further hypothesize that when healthy subjects wear the orthosis on both legs, we will observe symmetric kinematics and kinetics at the hip, knee, and ankle will be scaled with respect to that observed when people walk with no orthosis. We envisage the results of this pilot study on healthy subjects will provide an insight to the control parameters necessary when testing the active orthosis on stroke patients.

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II. AKROD v.2

AKROD (Active Knee Rehabilitation Orthotic Device) (Figure 1) is a programmable actuated knee orthosis for gait retraining and rehabilitation of stroke patients. It aims to foster training of more efficient and clinically desirable patterns of knee biomechanics during gait in stroke patients. The fully developed AKROD will have both resistive (variable dampers) and active (torque actuator) components. The resistive component facilitates knee flexion during stance by providing resistance to knee buckling while the active component encourages the patients to actively extend the knee during mid-to-terminal stance, facilitate knee flexion during initial swing, and again encourage knee extension during mid-to terminal stance. The variable damper is an electro-rheological fluid (ERF) based mechanism. ERF has a property of changing its viscosity when an electrical field is applied to it. Concentric cylinders, acting as electrodes supply the necessary electric field to activate the fluid, which changes its consistency from that of a fluid to a thick visco-elastic gel [6].

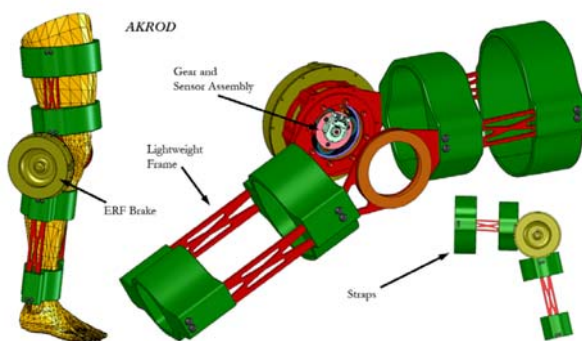


Figure 1: Schematic of the Active Knee Rehabilitation Orthotic Device (AKROD) with ERF brake.

While the AKROD is proposed as a tool to facilitate gait retraining in stroke patients, it must be acknowledged that the mass associated with the AKROD will likely cause gait adaptations. Although the AKROD can be set in a modality that makes it virtually neutral to the subject, and thus likely to cause very small adaptations in knee joint biomechanics, the additional weight introduced by the knee orthosis is expected to affect hip, knee and ankle kinematics. In this study, we intend to quantify gait responses of healthy subjects to wearing the additional mass of the AKROD, to study the speed dependence of these responses, and to explore the development of optimal strategies to minimize the biomechanical adaptations introduced by the knee orthosis.

III. METHODS

We tested one male subject who reported no neurological or musculoskeletal problems at the time of this pilot study. We asked the subject to ambulate along a level walkway at two speeds, $\sim 0.6 \text{ m}\cdot\text{s}^{-1}$ and $\sim 0.9 \text{ m}\cdot\text{s}^{-1}$

while wearing either (1) no orthosis, (2) an orthosis uni-laterally (right lower limb), or (3) an orthosis bi-laterally (left and right lower limbs). The orthoses used exhibited the same physical characteristics (i.e. design and mass distribution) as the AKROD but without the resistive and active components about the articulation of the orthosis. These components were replaced by machined stainless steel parts with the same shape and mass distribution (Figure 2). The orthosis was secured to each leg with velcro straps attached to the proximal and distal struts. Additionally, to minimize downward migration during the walking trials the orthosis was suspended proximally from a pelvic brace and supported distally with a plastic footplate under the heel attaching near the ankle. The order of the orthosis conditions was mixed while the two speeds for each condition were block randomized. The subject completed 10 walking trials for each orthosis-speed condition.

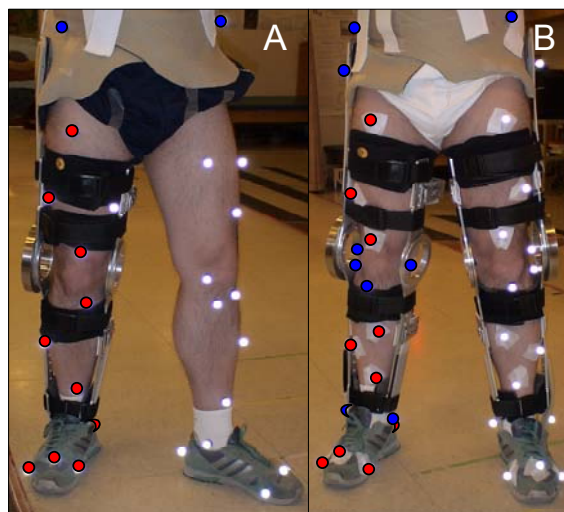


Figure 2: Technical (red) and anatomical (blue) markers of the right leg for the (A) uni-lateral orthosis condition, and (B) bi-lateral orthoses condition. Technical and anatomical markers were the same for the feet.

An 8-camera motion analysis system (Vicon 512, Vicon Peak, Oxford, UK) with two force platforms (AMTI, Watertown, MA) embedded in the walkway was used to collect kinematic and kinetic data for each lower limb during the walking trials. We will only report on kinematic data. Kinematics were described from the trajectories of reflective markers attached to the lower limbs of the subject. A set of “technical” marker clusters was attached to the skin over bony landmarks of the pelvis and each foot, and the anterior aspects of each thigh and shank (Figure 2). Additional “anatomical” markers were attached to specific anterior bony landmarks of the pelvis and proximal and distal bony landmarks of each femur, tibia and fibula before each block of walking trials for the respective conditions. The technical and anatomical markers were coincidental for the feet. The relative position and orientation of the “technical” marker clusters on the segments defined by the “anatomical” markers was recorded via a static standing calibration trial. The “anatomical” markers for each thigh and shank were then removed prior to the walking trials. Translation-rotation

matrices of the respective marker clusters defining each segment were used to quantify the kinematics of the hip, knee and ankle of each lower limb during the dynamic walking trials.

IV. RESULTS

Figure 3 shows the kinematics of sagittal motion at the hip, knee, and ankle in the three experimental conditions of level walking 1) with no orthosis, 2) with the orthosis on the right lower limb (uni-lateral), and 3) with two orthoses (bi-lateral). Data is shown for one subject walking at $\sim 0.9 \text{ m}\cdot\text{s}^{-1}$.

At the hip, very similar hip flexion/extension profiles are shown for the no orthosis condition and the bi-lateral orthoses condition. Significant deviations characterize instead the uni-lateral orthosis condition. During early stance, the hip profile for the uni-lateral condition is marked by a faster extension pattern than for the no orthosis and bilateral conditions. The derivative of the hip extension then decreases to almost reach zero for the uni-lateral condition, while it continues to decrease at an almost constant rate for the no orthosis and bilateral condition until late stance. Peak hip extension is about the same across conditions. In contrast, the peak hip flexion in terminal swing is slightly decreased for the uni-lateral

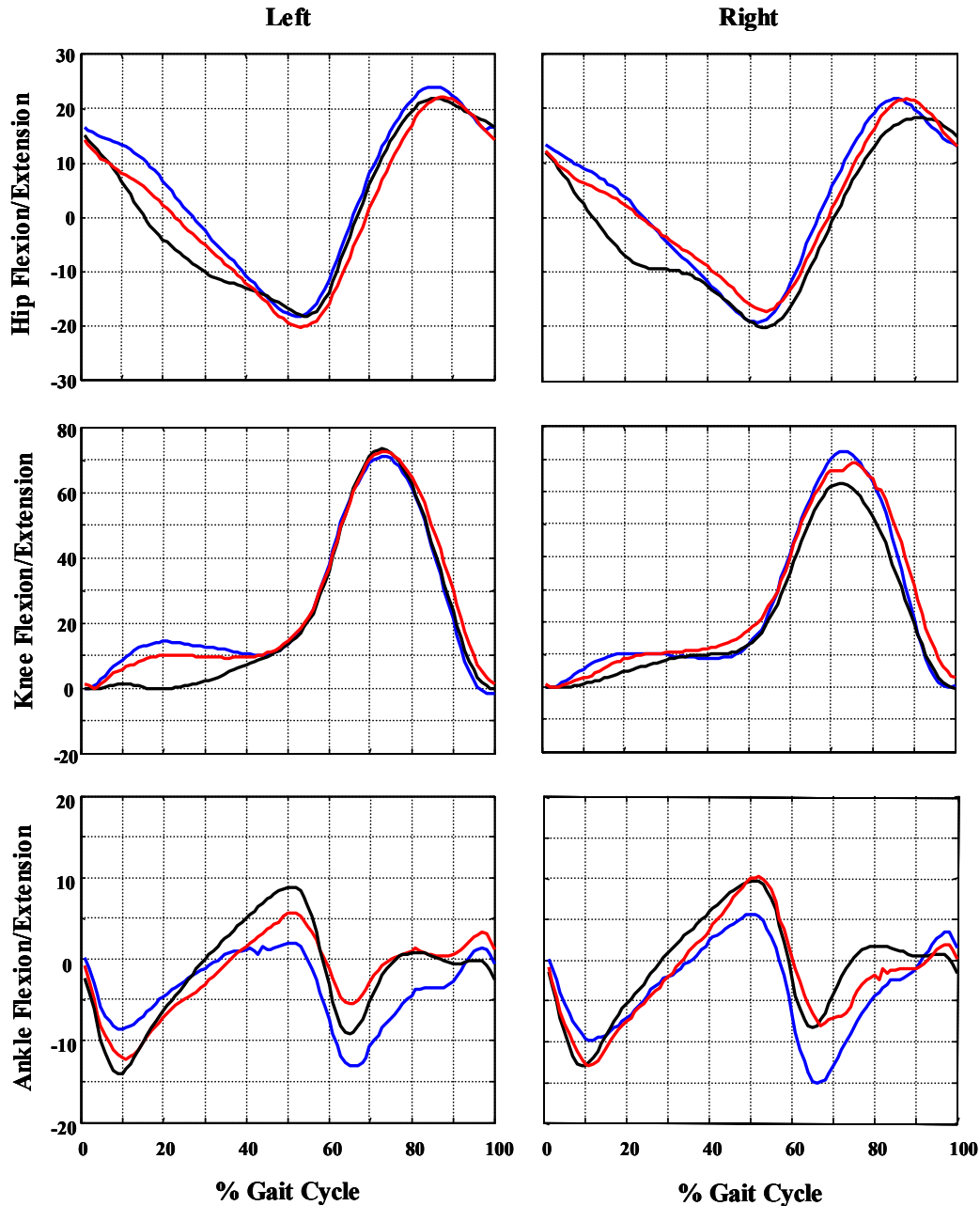


Figure 3: Mean sagittal motion kinematics of the left and right hip, knee and ankle for the 3 conditions: no orthosis (blue line), uni-lateral orthosis on the right side (black line), and (3) bi-lateral orthosis (red line) – for level walking at $\sim 0.9 \text{ m}\cdot\text{s}^{-1}$.

condition compared to the no orthosis and bi-lateral conditions.

At the knee, the bi-lateral condition shows a pattern that is overall closer to the no orthosis condition than the uni-lateral condition. During early to mid-stance, the knee flexion pattern (load absorption phase) for the bi-lateral condition is similar to that observed for the no orthosis condition. There is a decreased peak knee flexion during the swing phase for the uni-lateral condition compared to the other two conditions. In general, the results for the bi-lateral condition appear more symmetric than the data for the uni-lateral condition.

Finally, at the ankle, both the uni-lateral and the bi-lateral conditions show a decreased peak plantarflexion during early to mid-stance, an increase peak dorsiflexion during late stance, and a decreased peak plantarflexion during early swing compared to the no orthosis condition.

V. DISCUSSION

The adaptations observed in the kinematics of hip, knee, and ankle motion in the three experimental conditions (i.e. no orthosis, one orthosis (uni-lateral), and two orthoses (bi-lateral)) indicate that the control of the lower extremity movement pattern is most difficult when an asymmetric load is attached to the lower limbs, as in the case of the uni-lateral condition. The main adaptation pattern that characterizes the uni-lateral condition appears to be the lack of knee flexion during early to mid-stance. The stiff-leg pattern guarantees relative symmetry at the knee, but causes deviations of the hip and ankle kinematics. At the hip, the lack of knee flexion observed in the uni-lateral condition leads to an exaggerated hip flexion pattern in early to mid-stance. At the ankle, the stiff-leg pattern results in an exaggerated plantarflexion in early to mid-stance. Also, the mass associated with the brace appears to affect control of tibia progression during mid to terminal-stance thus leading to an exaggerated ankle dorsiflexion. Finally, the foot attachment at the ankle and heel appears to constrain ankle plantarflexion pattern around toe-off. All these adaptations appear to be significantly larger in the patterns associated with the uni-lateral condition compared to the bi-lateral condition. It follows that the use of two orthoses allows one to minimize the perturbation of the gait patterns caused by the use of the knee orthosis to an extent that appear to be acceptable for gait retraining purposes at slow walking speeds.

VI. CONCLUSION

The results of this pilot study indicate that the use of a robotic knee orthosis affects movement patterns during level walking at slow speeds. In the context of gait retraining, this observation points to the need for pursuing alternative strategies to avoid interfering with the re-learning of normal walking patterns. A very simple strategy based on a symmetric load of the lower extremity has been tested. Results indicate that this simple solution may be adequate as it effectively minimizes the adaptive

patterns observed when wearing one orthosis uni-laterally. Besides, joint kinematics in the bi-lateral condition appear quite similar to the results gathered when subjects walked with no orthosis.

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