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Mechanical performance of artificial pneumatic muscles to power an ankle–foot orthosis

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Abstract

We developed a powered ankle–foot orthosis that uses artificial pneumatic muscles to produce active plantar flexor torque. The purpose of this study was to quantify the mechanical performance of the orthosis during human walking. Three subjects walked at a range of speeds wearing ankle–foot orthoses with either one or two artificial muscles working in parallel. The orthosis produced similar total peak plantar flexor torque and network across speeds independent of the number of muscles used. The orthosis generated ~57% of the peak ankle plantar flexor torque during stance and performed ~70% of the positive plantar flexor work done during normal walking. Artificial muscle bandwidth and force–length properties were the two primary factors limiting torque production. The lack of peak force and work differences between single and double muscle conditions can be explained by force–length properties. Subjects altered their ankle kinematics between conditions resulting in changes in artificial muscle length. In the double muscle condition greater plantar flexion yielded shorter artificial muscles lengths and decreased muscle forces. This finding emphasizes the importance of human testing in the design and development of robotic exoskeleton devices for assisting human movement. The results of this study outline the mechanical performance limitations of an ankle–foot orthosis powered by artificial pneumatic muscles. This orthosis could be valuable for gait rehabilitation and for studies investigating neuromechanical control of human walking.

Keywords: Locomotion; Exoskeleton; Gait; Rehabilitation; Work

1. Introduction

A powered ankle–foot orthosis could be very useful for basic science studies investigating the neuromechanical control of human walking. We are particularly interested in powered plantar flexion because the plantar flexor muscles are critical to the generation of forward velocity and support of the center of mass during human walking (Gottschall and Kram, 2003; Kepple et al., 1997; Meinders et al., 1998; Neptune et al., 2001; Winter, 1983). Robotic manipulanda for the upper limb have provided evidence that the nervous systems builds internal models of limb dynamics and have revolutionized our understanding of the neural control of human arm movements (Bizzi and Mussa-Ivaldi, 1998; Kawato, 1999; Mussa-Ivaldi and Bizzi, 2000; Reinkensmeyer et al., 2004; Scheidt et al., 2000; Shadmehr and Moussavi, 2000; Shadmehr and Mussa-Ivaldi, 1994). For example, an initial study found that humans plan upper limb movements in joint space coordinates rather than end pont coordinates (Shadmehr and Mussa-Ivaldi, 1994). Another study found motor adaptation with one arm transfers to the other arm (Criscimagna-Hemminger et al., 2003). In contrast to the many studies

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utilizing robotic devices to study upper limb movement, there have been very few that employ similar techniques to investigate lower limb movements (Reinkensmeyer et al., 2004). Using robotic devices to perturb lower limb movements during gait may give insights into how the human nervous system governs locomotion. A powered orthosis could be used to perform analogous studies to those performed on the upper limb to investigate movement planning, learning transfer and internal modeling during locomotion. As a result, powered lower limb orthoses would be valuable for investigating neuromechanical control of human walking.

A powered ankle–foot orthosis would also be valuable for gait rehabilitation purposes. Breakthroughs in clinical neuroscience have revealed that humans with spinal cord injury or stroke can increase their motor capabilities through intense task-specific practice (Barbeau et al., 1998; Dietz et al., 1998; Harkema, 2001; Hesse et al., 1995; Wernig et al., 1995). Manual assistance from several physical therapists is often required for therapy (Behrman and Harkema, 2000). Robotic devices could substantially reduce manual labor costs. Several groups are working on robotic devices to aid gait rehabilitation after neurological injury (Colombo et al., 2000; Edgerton et al., 2001; Hesse and Uhlenbrock, 2000; Hesse et al., 2000; Jezernik et al., 2003; Reinkensmeyer, 2003). However, none of these devices specifically provide plantar flexor assistance during walking. Two powered ankle–foot orthoses have been described in the literature (Andersen and Sinkjaer, 1995, 2003; Blaya and Herr, 2004), but both have only been used to provide dorisflexion torque, not plantar flexion assistance.

We have developed a simple lightweight powered ankle–foot orthosis that can provide plantar flexion assistance during walking (Ferris et al., 2005, in press). The orthosis is actuated by artificial pneumatic muscles (i.e. Mckibben muscles or flexible pneumatic actuators). The artificial pneumatic muscles consist of an expandable internal bladder surrounded by a braided shell. When the internal bladder is pressurized, it expands in a balloon-like manner. The braided shell constrains the expansion. As the volume of the internal bladder increases with increasing pressure, the pneumatic muscle shortens and/or produces tension if coupled to a mechanical load. Artificial pneumatic muscles are desirable for our application because they are lightweight, capable of high forces, and inherently compliant (Davis et al., 2003; Klute et al., 2002; Reynolds et al., 2003; Tondu and Lopez, 2000). Their mechanical properities (force–length, force–velocity, force–pressure relationships) have been described in detail during benchtop testing (Klute et al., 1999; Klute and Hannaford, 1998, 2000). While the mechanics of artificial pneumatic muscles are clear, it is unpredictable how humans will respond mechanically when walking with a powered orthosis. The kinematics that arise from this human–machine interaction will regulate the force output of the artificial muscles.

The purpose of this study is to quantify the mechanical performance of artificial pneumatic muscles as they assist plantar flexion during human walking. One objective was to examine the effect of walking speed on the peak torque and work performed by the artificial pneumatic muscles. Walking speed may be a factor for two reasons. First, artificial pneumatic muscles have inherent bandwidth limitations (Davis et al., 2002). The time required for the artificial muscle to fully inflate, and produce maximal force is nearly constant. As walking speed increases subjects move faster through their ankle range of motion. It is possible that as walking speed increases peak torque and work performed may decrease because the artificial muscle length will be shorter when the artificial muscle reaches maximum inflation. Second, at faster walking speeds, there is a larger range of motion of the joint ankle that will presumably alter artifical pneumatic muscle length. A second objective was to quantifiy the advantages of using multiple artificial muscles instead of a single muscle. Using multiple artificial muscles in parallel (or increasing the cross-sectional area of a single artificial muscle) will increase peak isometric forces, but it is not clear how this will affect the mechanics of human walking.

2. Methods

Three healthy male subjects (height 1.80 ± 0.01 m; body mass 96 ± 9 kg; 30 ± 5 years of age; mean$\pm$s.d.) gave written informed consent and participated in this study. The protocol was approved by the University of Michigan Medical School Institution Review Board for Human Subject Research.

We custom fit each subject with an ankle–foot orthosis (Fig. 1) for their left lower limb. Construction of earlier prototypes of the ankle–foot orthosis have been described in detail (Ferris et al., 2005, in press). The orthosis consisted of a carbon fiber shank section and a polypropylene foot section. A metal hinge between the shank and foot sections allowed free sagittal rotation of the ankle joint. We attached either one or two artificial pneumatic muscles to the posterior of the orthosis. We connected a tension/compression force transducer in series with each artificial muscle. The total weight of the ankle–foot orthosis was 1.3–1.7 kg (Table 1) and had an average moment arm length of 10.1 ± 0.9 cm (mean$\pm$s.d.). Four parallel proportional pressure regulators (MAC Valves, Inc., Wixom, MI) supplied compressed air (0–6.2 bar) to each artificial muscle via nylon tubing. We attached an analog-controlled solenoid valve (MAC Valves, Inc., Wixom, MI) in parallel

Fig. 1. Powered ankle–foot orthosis. Each ankle–foot orthosis consists of a custom fit carbon fiber shank section and a polypropylene foot section. Load sensors in series with the artificial muscles monitored artificial muscle tension. (A) An orthosis fit with a single artificial pneumatic muscle to provide plantar flexion assistance. (B) The same orthosis fit with two artificial pneumatic muscles in parallel to provide plantar flexion assistance. All artificial muscles pictured are in a relaxed state.

Table 1
Ankle–foot orthosis measurements

Components	Weight (g)			
	Single muscle		Double muscle	
	Mean	SD	Mean	SD
Artificial muscle(s)	147	6	294	12
Load cell (s)	85	0	170	0
Shank section	768	22	794	25
Foot section	374	25	428	25
Total	1374	53	1686	62

Mean and standard deviations of individual component weights (g) of the three ankle–foot orthoses.

with the air supply tubing to each muscle to facilitate exhaust.

We used a real time computer interface (dSPACE Inc., Northville, MI; 1000 Hz) to control air pressure supplied to the artificial pneumatic muscles based on foot contact with the ground. The subject wore a foot switch, designed to fit only under their left forefoot inside their shoe. When the forefoot was in contact with the ground, a control signal was sent to the pressure regulators to activate maximal air pressure to the artificial pneumatic muscle(s).

We calculated isometric force–length properties and bandwidth of the artificial muscles during benchtop tests. We characterized bandwidth by exciting the system with sinusoidal inputs at a range of frequencies and computing the magnitude ratio of the output force amplitude to input voltage amplitude. The bandwidth was defined as the frequency at which the frequency response declined 3 db from its low-frequency value (Ogata, 2002).

During treadmill walking trials we collected bilateral joint angles, foot–ground contact, and artificial muscle force and length. We recorded bilateral ankle, knee and hip angles using electrogoniometers (Biometrics, Ltd, Ladysmith, VA). We recorded step cycle data using a pair of complete footswitches (B & L Engineering, Tustin, CA) placed in each shoe. We collected all analog data at 1200 Hz. We recorded the length of the artificial pneumatic muscle by attaching a reflective marker to each end of the muscle and tracking its motion using a six camera kinematic system (Motion Analysis Corporation, Santa Rosa, CA; 120 Hz).

The subjects walked on the treadmill at four different speeds (0.5, 1.0, 1.5 and 2.0 m/s) and at five different conditions per speed. The five walking conditions were: normal (*no orthosis*), ankle–foot orthosis with a single passive muscle attached (*single passive*), ankle–foot orthosis with a single active muscle attached (*single active*), ankle–foot orthosis with two passive muscles attached (*double passive*), and ankle–foot orthosis with two active muscles attached (*double active*). One subject repeated the single passive and active conditions at 1.0 m/s with artificial muscles of three different maximal lengths (45, 46 and 47 cm) to determine the effect of artificial muscle length. At every condition, the subjects walked until they felt comfortable (approximately 1 min) before we collected data. We recorded 20 s of data for each condition (between 8 and 20 gait cycles depending on speed). Every condition was repeated two times during a testing session in quasi-randomized order.

We recorded data during over ground walking at 1.0 m/s for the single passive and active conditions. During over ground walking, we recorded ground reaction force data from two force plates in addition to all data recorded during treadmill walking except electrogoniometer data. We used nine reflective markers placed on the left shank and foot to calculate lower limb kinematics for over ground trials. A stopwatch connected to two light triggers was used to determine over ground walking speed. We collected 10 trials for both the passive and active conditions. Prior to each over ground condition, the subject walked on a treadmill for 5 min to become accommodated to the condition.

We calculated net moments about the lower limb joints using commercial software (Visual3D, C-Motion, Inc., Rockville, MD) combining kinematic marker and force platform data. Lower limb inertial properties were estimated based on anthropometric measurements of the subjects (Zatsiorsky, 2002). For the inverse dynamics calculations, we modified segment inertial parameters of the lower limb to account for orthosis mass and moment of inertia.

We calculated artificial muscle work from the artificial muscle force and length data. We also calculated orthosis work from the artificial muscle torque and ankle angle data. Differences in artificial muscle work and orthosis work could indicate energy loses.

We used six repeated measures ANOVAs to test for differences in peak artificial muscle force, net muscle work, positive muscle work, negative muscle work, positive orthosis work and negative orthosis work between walking speeds, and between single vs. double orthosis conditions. We also used two repeated measures ANOVAs to test for differences in positive work and negative work between that performed by the artificial muscle(s) and orthosis (JMP IN software, SAS Institute, Inc.).

3. Results

During isometric benchtop testing, the artificial pneumatic muscles demonstrated a linear force–length relationship (Fig. 2). A single artificial muscle produced a peak force of 1700 N when fully activated at its maximal length. Force in the muscle decreased to zero when contracted to 71% of its maximum length. When two artificial muscles were placed in parallel under isometric conditions, the total force produced by two muscles was double the force produced by a single muscle. The bandwidth of the muscle determined from benchtop testing was 2.4 ± 0.1 Hz (mean $\pm$ s.d.).

The subjects had similar ankle, knee and hip kinematics at a given speed during normal, single passive and double passive conditions (Fig. 3A). During passive trials, the artificial muscles created small forces (<40 N) at maximum ankle dorsiflexion because the muscle(s) was/were passively stretched (Fig. 3A).

During active conditions, the artificial pneumatic muscles produced large forces during stance and performed substantial positive work. Total force produced by the artificial muscles was similar for single and double active conditions and across speeds (Figs. 3B, 4A and 5). Results of the ANOVA comparing peak artificial

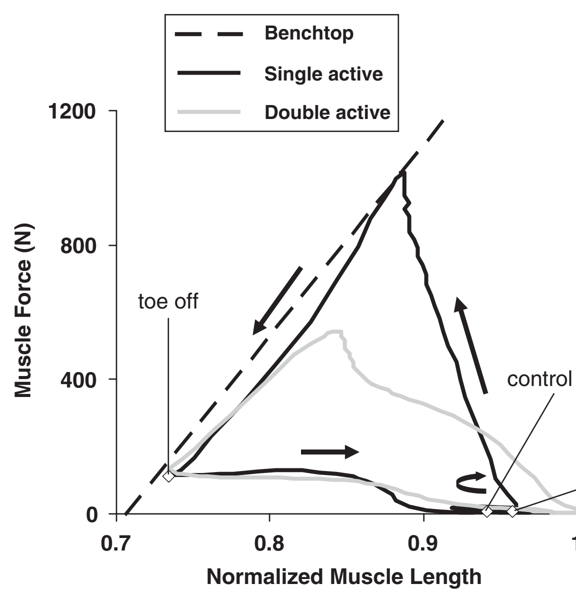

Fig. 2. Force–length relationship of artificial pneumatic muscles. Dashed line is the force–length relationship of an artificial pneumatic muscle during isometric benchtop testing. Although not pictured, the muscle will achieve a peak force of 1700 N at maximal length (1.0) during isometric testing. Single active (solid black line) is mean force–length data recorded from the artificial pneumatic muscle during the active single condition. Double active (gray line) is mean force–length data recorded from one of the two artificial pneumatic muscles working in parallel during the double active condition. Data for the single and double active conditions are representative data from one subject walking at 1.0 m/s. Arrows represent direction of the force–length changes. Heel strike, toe off and control signal onset for the active single condition are marked on the force–length curve.

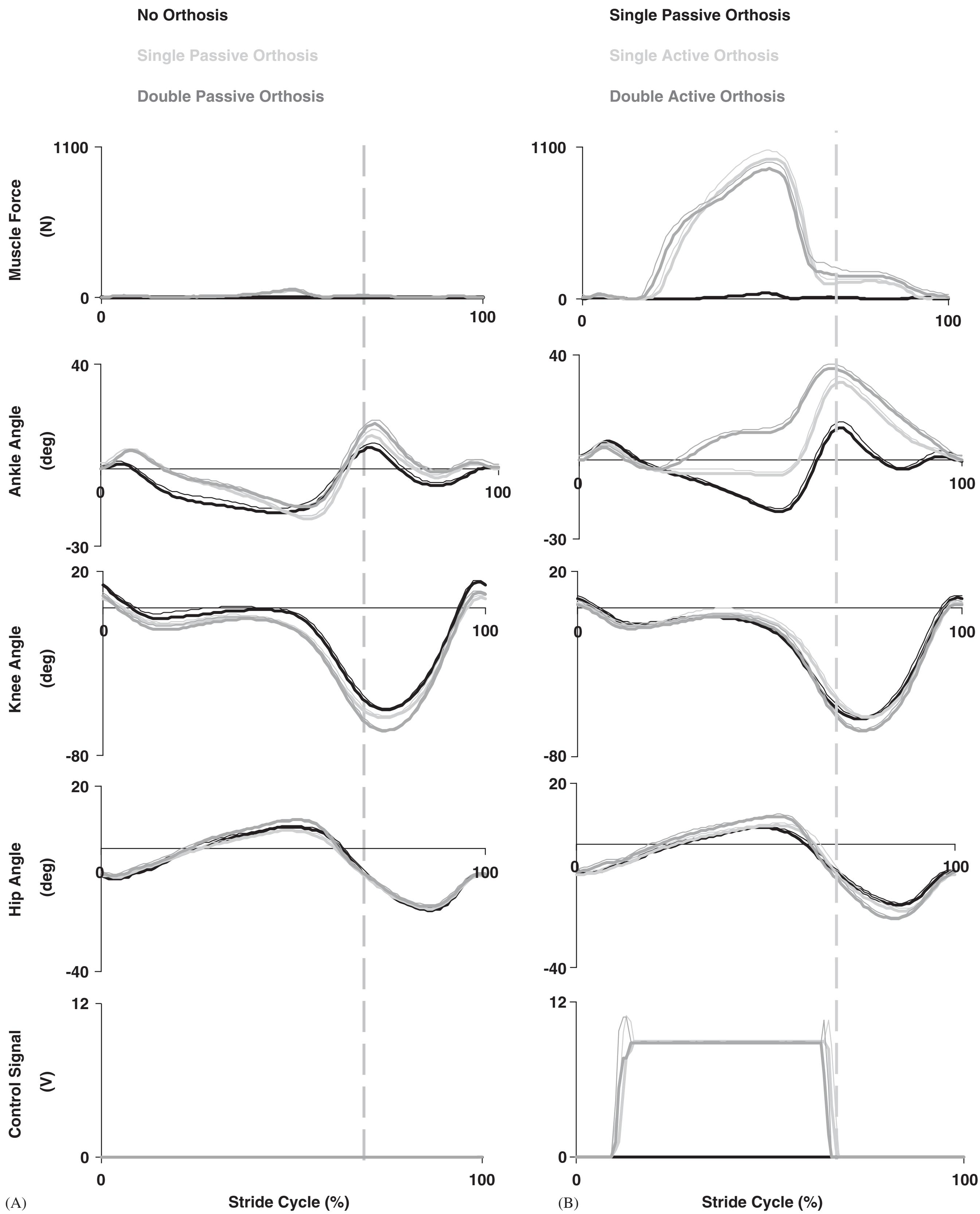

Fig. 3. Muscle force, joint kinematics and control signal. Artificial muscle force, left joint kinematics and control signal for one subject walking at 1.0 m/s (mean + standard deviation). Data are normalized from left heel strike to left heel strike. Dashed line indicates toe off. (A) No othosis, single passive and double passive conditions. (B) Single passive, single active and double active conditions. We defined standing posture as zero degrees for joint angles. Positive is plantar flexion for the ankle, extension for the knee and hip joint. Zero control signal is a command to release all pressure from the artificial muscles. A 10 V control signal is a command to send maximal air pressure to the artificial muscle(s).

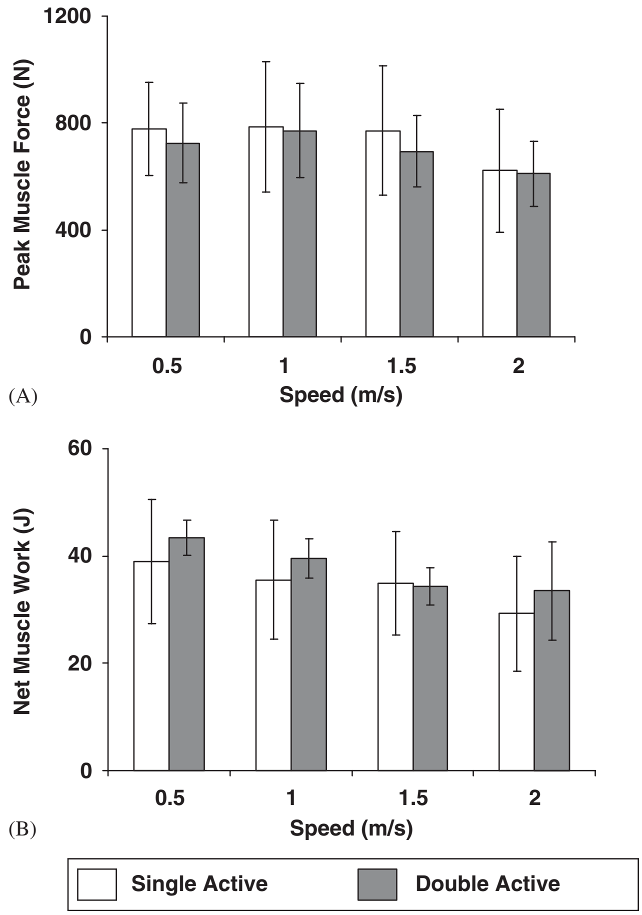

Fig. 4. Peak force and net work. (A) Mean and standard deviation of peak artificial muscle force during a gait cycle for the three subjects. Data from the double active condition are the sum of the forces in both muscles. (B) Mean and standard deviation of net muscle work performed by the artificial muscles during a gait cycle for the three subjects. Data from the double active condition are the sum of net work performed by each muscle.

muscle force revealed no significant difference between single and double muscle conditions ($F_{(1,17)} = 0.7191$, $p = 0.4082$), or speed ($F_{(3,17)} = 2.4627$, $p = 0.0976$). Net artificial muscle work performed was also similar between single and double active conditions and across speeds (Fig. 4B). ANOVA results found no difference by speed or single vs. double conditions in net muscle work (total model: $F_{(6,17)} = 2.6324$, $p = 0.0543$). A power analysis (Sall et al., 2001) revealed that the effect sizes for single vs. double conditions were small (force 0.13, work 0.18). Testing 79 subjects would produce a 50% chance of detecting real differences between single and double conditions if they existed. Thus, if there were any true differences between conditions, they were not great. The effect sizes for speed were larger (force 0.51, work 0.48), but we would still need at least 30 subjects to have a greater than 50% chance of detecting real differences in speed if they existed.

Similarly, ANOVA results found no significant differences in positive muscle work, negative muscle work or positive orthosis work between single vs. double conditions or across speeds ($p > 0.05$) (Table 2). A difference in negative orthosis work was found between single vs. double conditions ($F_{(1,17)} = 13.5936$, $p = 0.0018$) but not across speed ($F_{(3,17)} = 0.03225$, $p = 0.8090$) (Table 2).

Positive and negative work performed by the artificial muscle(s) were significantly greater than that performed by the orthosis (positive work: $F_{(3,44)} = 15.1266$, $p < 0.0001$) (negative work: $F_{(3,44)} = 10.5090$, $p < 0.0001$) (Table 2). Artificial muscle work and orthosis work are not necessarily the same. Compliance in the muscle attachment brackets and orthosis shell can result in artificial muscle shortening without concomitant changes in joint angle.

Subjects walked with greater plantar flexion during active trials compared to normal and passive trials (Figs. 3B and 5). Kinematics at the hip and knee were not noticeably different during active trials compared to normal and passive trials. The differences in ankle kinematics between conditions resulted in differences in artificial muscle length. As subjects increased plantar flexion, the artificial muscles had shorter lengths (Figs. 3B and 5).

A plot of artificial muscle force vs. length during the single active condition shows typical mechanical behavior during walking (Fig. 2). At initial heel strike, the force in the artificial muscle was zero. From heel strike to peak artificial muscle force, the artificial muscle was inflating. Consequently, muscle bandwidth limited force development. Between peak force and toe off, force decreased as the muscle shortened. Muscle force during this time period was almost completely dictated by the force–length relationship of the muscle. At toe off, the control signal shut off but there was a mechanical delay in relaxation caused while air released from the muscle. The subject dorsiflexed during swing, increasing muscle length as air was being released. The force returned to zero before heel strike and then the cycle repeated. This force vs. length activation profile demonstrates that torque produced by the artificial muscle was dictated by both bandwidth and force–length properties.

A major difference between single and double active conditions was that the subject walked more plantar flexed during the double condition compared to the single condition. This resulted in the artificial muscles producing force at shorter lengths (Fig. 2). The peak force produced by the single muscle was similar to the peak force produced by the two parallel muscles because of the difference in muscle lengths.

There was an optimal maximal length for the artificial muscle (Fig. 6). If the muscle was too long, peak force was decreased when it was active. If the muscle was too short, the muscle developed high passive forces.

Subjects demonstrated similar net ankle joint moments during both passive and active conditions (Fig. 7).

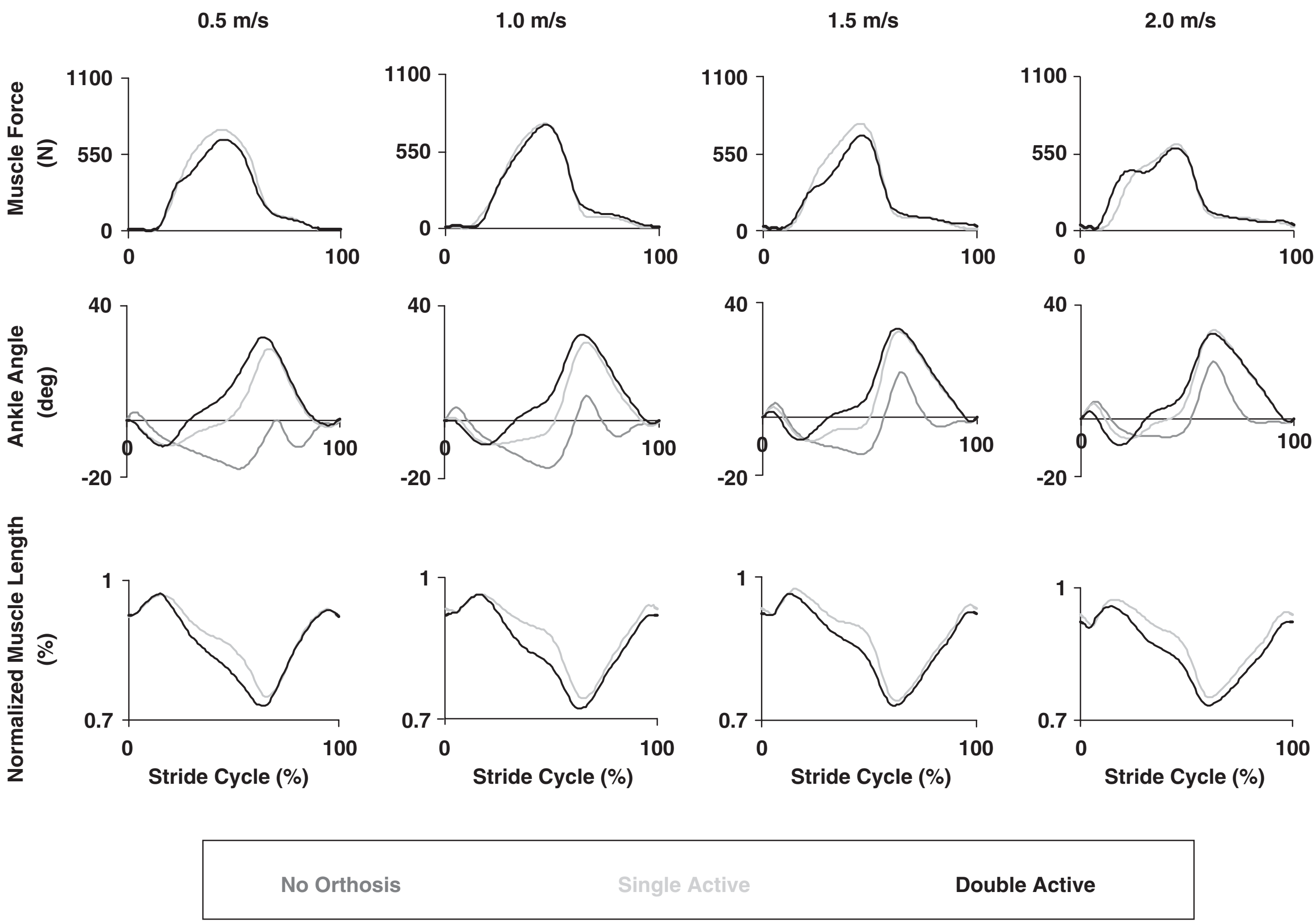

Fig. 5. Mean muscle force, muscle length and ankle angle. Mean artificial muscle force, muscle length and ankle angle from the three subjects at four different speeds during the no orthosis, single active and double active conditions. Data are normalized from left heel strike to left heel strike. Ankle plantar flexion is positive. Zero is neutral ankle angle. 1.0 is maximal artificial muscle length.

Table 2
Artificial muscle and orthosis work

Speed (m/s)		Work (J)							
		Single active				Double active			
		Muscle		Orthosis		Muscle		Orthosis	
		Mean	SD	Mean	SD	Mean	SD	Mean	SD
0.5	Pos.	45.0	9.4	28.0	5.5	49.9	3.0	30.8	14.3
	Neg.	6.1	3.3	0.2	0.1	6.5	4.2	3.8	2.6
1.0	Pos.	40.7	10.8	26.4	5.2	46.3	6.3	35.0	4.9
	Neg.	5.1	2.9	0.9	1.2	6.8	3.3	4.9	2.6
1.5	Pos.	40.6	9.9	27.3	3.7	41.0	7.2	34.1	7.3
	Neg.	5.6	0.6	0.9	1.1	6.6	3.8	5.9	7.0
2.0	Pos.	35.6	12.9	22.2	7.5	40.0	9.7	27.2	5.4
	Neg.	6.3	2.2	1.2	1.1	6.5	0.8	3.7	3.5

Mean and standard deviations of positive and negative work performed by the artificial muscle(s) and the orthosis during the gait cycle. Data are from single and double active conditions for all three subjects. Muscle work was calculated directly from artificial muscle force and length data. Orthosis work was calculated from artificial muscle torque and ankle angle.

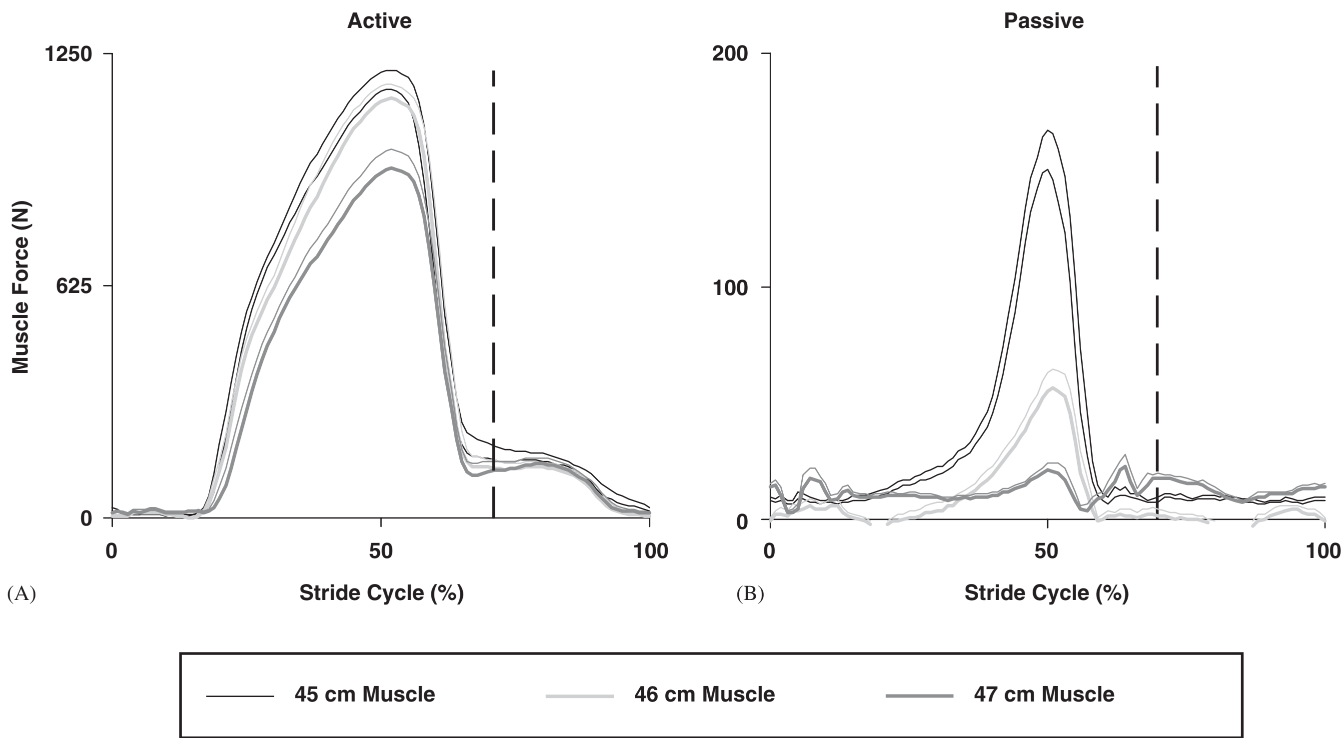

Fig. 6. Maximal muscle length vs. force. Mean + standard deviation of artificial muscle force data from one subject walking at 1.0 m/s during the (A) single active and (B) single passive conditions. Conditions were repeated three times with three different artificial pneumatic muscles fit in the orthosis. Each muscle had a different maximal length (45, 46, 47 cm).

The artificial muscle produced a peak plantar flexor moment that was 57% of the net ankle plantar flexor moment during stance. The subjects' net ankle moment was zero during swing in spite of the passive plantar flexor moment produced by the artificial muscle.

4. Discussion

In the ankle–foot orthosis design we tested, the artificial pneumatic muscles were able to substantially assist plantar flexion during walking. At 1.0 m/s a single artificial pneumatic muscle generated peak torque that was 57% of the maximum ankle plantar flexor torque during stance. The orthosis peak torque did not change with walking speed. The orthosis produced 0.28 J/kg of positive work at 1.0 m/s and 0.31 J/kg of positive work at 1.5 m/s. These values are about 70% of the positive plantar flexor work done during normal walking (Eng and Winter, 1995). Artificial muscle work also remained fairly constant across speeds. Across all conditions, the actual positive work performed by the artificial muscle was greater than the positive work performed by the orthosis about the ankle. This discrepancy in work is likely a result of energy loses when transferring power from the artificial muscle to the orthosis. These loses

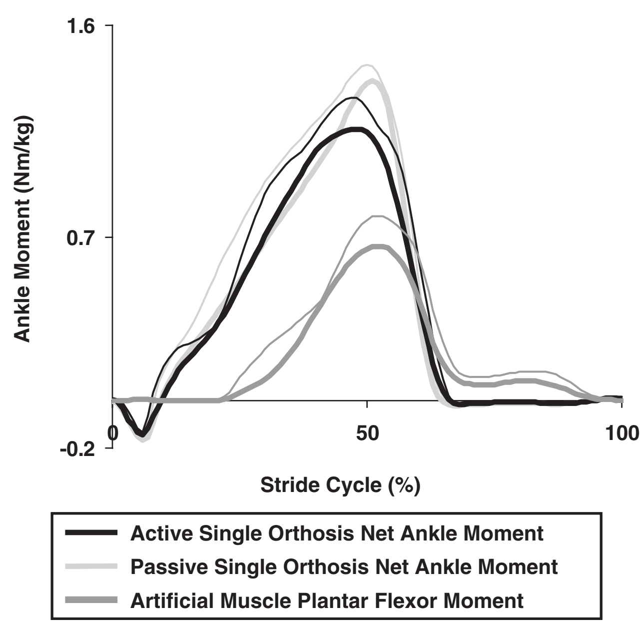

Fig. 7. Net ankle and artificial muscle moments. Mean + standard deviation of net ankle moments and artificial pneumatic muscle moments for the three subjects during over ground walking at 1.0 m/s. Net ankle moments are during the single passive and single active conditions. Artificial pneumatic muscle moments are from the single active condition. Plantar flexion moments are in the positive direction.

could be caused by small deflections of the orthosis at the steel attachment brackets, the polypropylene foot or carbon fiber shank sections. Increasing the rigidity of future orthoses may reduce energy loses.

The mechanical performance of the artificial muscles was partially limited by their bandwidth. In spite of the bandwidth limitations, peak muscle force and muscle work were not greatly affected by walking speed. The artificial pneumatic muscle bandwidth of 2.4 Hz was actually very similar to the bandwidth of human muscle ~2.2 Hz (Aaron and Stein, 1976).

There are numerous methods to alter the bandwidth of artificial pneumatic muscles (Davis et al., 2003). Two common methods to increase bandwidth are by decreasing dead space volume or by increasing flow rate. Decreasing muscle size (e.g. length or cross-sectional area) or placing filler inside the muscle will decrease dead space volume. Changing the type or increasing the number or pressure regulators in parallel can increase flow rate. Davis et al. (2003) offer a comprehensive review of factors determining artificial pneumatic muscle bandwidth.

Our findings were that artificial pneumatic muscle forces in our orthosis were largely dictated by the force–length relationship. Once the artificial muscles were fully inflated, the force was close to the isometric peak force of the muscle at that length. Future orthosis designs could minimize the effect of muscle length on muscle force by moving muscle attachments closer to joint centers. However, this would decrease moment arm lengths and result in less torque for a given muscle force. Alternatively, future orthosis designs could use longer artificial muscles to produce active forces over a greater range of joint motion. Increasing artificial muscle length is limited due to the geometric limitations of the human lower limb.

The lack of peak force and work differences between single and double muscle conditions can be explained by force–length properties. During the initial part of stance, total artificial muscle force was greater for the double condition than the single condition. The greater torque, accelerated the ankle joint into plantar flexion. This change in kinematics resulted in a shorter artificial muscle length later in stance for the double condition compared to the single condition. At the point of peak torque the artificial muscles for the double condition produced less force because they were shorter than in the single condition. This finding emphasizes the importance of human testing in the design and development of robotic exoskeleton devices for assisting human movement.

Two other powered ankle–foot orthoses have previously been described (Andersen and Sinkjaer, 1995, 2003; Blaya and Herr, 2004). Both have only been used to provide dorsiflexor torque during walking. Our ankle–foot orthosis is unique in providing plantar flexion torque during walking. There are also differences in the type of actuators used to create torque. Blaya and Herr (2004) used an electromechanical series elastic actuator. The advantages of this actuator include increased bandwidth and no force–length dependency (Pratt et al., 2002). The disadvantage is a much greater weight (Pratt et al., 2002). Andersen and Sinkjaer (1995) created dorsiflexor torque using an ac-motor. Torque from the motor was transferred to the ankle using bowden wires. An advantage of electromechanical motors is the ability to produce large torques. However, electromechanical motors also have considerable impedance, which limits their backdrivability (Reinkensmeyer et al., 2004). A goal of our orthosis was to mimic natural gait movements. The low compliance and high backdrivability of the artificial pneumatic muscles used in our orthosis make them well suited for this purpose (Klute et al., 2002; Klute and Hannaford, 2000).

The results of this study outline the mechanical performance limitations of a powered ankle–foot orthosis design. We do not intend the orthosis to ever be used in a portable situation (e.g. to aid a patient walking in the community). The value of the orthosis will be for studies examining the biomechanics of human walking in a laboratory or clinic setting. Future research utilizing different control algorithms (e.g. proportional myoelectric or artificial neural oscillators) could be valuable for gaining insights into human neuromechanical control of locomotion.

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References

Aaron, S.L., Stein, R.B., 1976. Comparison of an EMG-controlled prosthesis and the normal human biceps brachii muscle. American Journal of Physical Medicine 55, 1–14.

Andersen, J.B., Sinkjaer, T., 1995. An actuator system for investigating electrophysiological and biomechanical features around the human ankle joint during gait. IEEE Transactions on Rehabilitation Engineering 3, 299–306.

Andersen, J.B., Sinkjaer, T., 2003. Mobile ankle and knee perturbator. IEEE Transactions on Biomedical Engineering 50, 1208–1211.

Barbeau, H., Norman, K., Fung, J., Visintin, M., Ladouceur, M., 1998. Does neurorehabilitation play a role in the recovery of walking in neurological populations? Annals of the New York Academy of Sciences 860, 377–392.

Behrman, A.L., Harkema, S.J., 2000. Locomotor training after human spinal cord injury: a series of case studies. Physical Therapy 80, 688–700.

Bizzi, E., Mussa-Ivaldi, F.A., 1998. Neural basis of motor control and its cognitive implications. Trends in Cognitive Sciences 2, 97–102.

Blaya, J.A., Herr, H., 2004. Adaptive control of a variable-impedance ankle–foot orthosis to assist drop-foot gait. IEEE Transactions on Neural Systems and Rehabilitation Engineering 12, 24–31.

Colombo, G., Joerg, M., Schreier, R., Dietz, V., 2000. Treadmill training of paraplegic patients using a robotic orthosis. Journal of Rehabilitation Research and Development 37, 693–700.

Criscimagna-Hemminger, S.E., Donchin, O., Gazzaniga, M.S., Shadmehr, R., 2003. Learned dynamics of reaching movements generalize from dominant to nondominant arm. Journal of Neurophysiology 89, 168–176.

Davis, S., Canderle, J., Artrit, P., Tsagarakis, N., Caldwell, D.G., (2002). Enhanced dynamic performance in pneumatic muscle actuators. *IEEE International Conference on Robotics and Automation*, Washington, DC, USA, pp. 2836–2841.

Davis, S., Tsagarakis, N., Canderle, J., Caldwell, D.G., 2003. Enhanced modelling and performance in braided pneumatic muscle actuators. International Journal of Robotics Research 22, 213–227.

Dietz, V., Wirz, M., Curt, A., Colombo, G., 1998. Locomotor pattern in paraplegic patients: training effects and recovery of spinal cord function. Spinal Cord 36, 380–390.

Edgerton, V.R., Leon, R.D., Harkema, S.J., Hodgson, J.A., London, N., Reinkensmeyer, D.J., Roy, R.R., Talmadge, R.J., Tillakaratne, N.J., Timoszyk, W., Tobin, A., 2001. Retraining the injured spinal cord. Journal of Physiology (London) 533, 15–22.

Eng, J.J., Winter, D.A., 1995. Kinetic analysis of the lower limbs during walking: what information can be gained from a three-dimensional model? Journal of Biomechanics 28, 753–758.

Ferris, D.P., Czerniecki, J.M., Hannaford, B., 2005. An ankle–foot orthosis powered by artificial pneumatic muscles. Journal of Applied Biomechanics 21, 189–197.

Ferris, D.P., Gordon, K.E., Sawicki, G.S., Peethambaran, A. An improved powered ankle–foot orthosis using proportional myoelectric control. Gait and Posture. In press.

Gottschall, J.S., Kram, R., 2003. Energy cost and muscular activity required for propulsion during walking. Journal of Applied Physiology 94, 1766–1772.

Harkema, S.J., 2001. Neural plasticity after human spinal cord injury: application of locomotor training to the rehabilitation of walking. Neuroscientist 7, 455–468.

Hesse, S., Bertelt, C., Jahnke, M.T., Schaffrin, A., Baake, P., Malezic, M., Mauritz, K.H., 1995. Treadmill training with partial body weight support compared with physiotherapy in nonambulatory hemiparetic patients. Stroke 26, 976–981.

Hesse, S., Uhlenbrock, D., 2000. A mechanized gait trainer for restoration of gait. Journal of Rehabilitation Research and Development 37, 701–708.

Hesse, S., Uhlenbrock, D., Werner, C., Bardeleben, A., 2000. A mechanized gait trainer for restoring gait in nonambulatory subjects. Archives of Physical Medicine and Rehabilitation 81, 1158–1161.

Jezernik, S., Scharer, R., Colombo, G., Morari, M., 2003. Adaptive robotic rehabilitation of locomotion: a clinical study in spinally injured individuals. Spinal Cord 41, 657–666.

Kawato, M., 1999. Internal models for motor control and trajectory planning. Current Opinion in Neurobiology 9, 718–727.

Kepple, T.M., Siegel, K.L., Stanhope, S.J., 1997. Relative contributions of the lower extremity joint moments to forward progression and support during gait. Gait and Posture 6, 1–8.

Klute, G.K., Czerniecki, J.M., Hannaford, B., 1999. McKibben artificial muscles: pneumatic actuators with biomechanical intelligence. *IEEE/ASME International Conference on Advanced Intelligent Mechatronics*, Atlanta, GA, pp. 221-226.

Klute, G.K., Czerniecki, J.M., Hannaford, B., 2002. Artificial muscles: actuators for biorobotic systems. International Journal of Robotics Research 21, 295–309.

Klute, G.K., Hannaford, B., 1998. Fatigue characteristics of McKibben artificial muscle actuators. *IEEE/RSJ International Conference on Intelligent Robots and Systems*, Victoria, BC, Canada, pp. 1776–1782

Klute, G.K., Hannaford, B., 2000. Accounting for elastic energy storage in McKibben artificial muscle actuators. Journal of Dynamic Systems, Measurement and Control 122, 386–388.

Meinders, M., Gitter, A., Czerniecki, J.M., 1998. The role of ankle plantar flexor muscle work during walking. Scandinavian Journal of Rehabilitation Medicine 30, 39–46.

Mussa-Ivaldi, F.A., Bizzi, E., 2000. Motor learning through the combination of primitives. Philosophical Transactions of the Royal Society of London Series B-Biological Sciences 355, 1755–1769.

Neptune, R.R., Kautz, S.A., Zajac, F.E., 2001. Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. Journal of Biomechanics 34, 1387–1398.

Ogata, K., 2002. Modern Control Engineering. Prentice-Hall, Upper Saddle River, NJ.

Pratt, J., Krupp, B., Morse, C., 2002. Series elastic actuators for high fidelity force control. Industrial Robot 29, 234–241.

Reinkensmeyer, D.J., 2003. Rehabilitators. In: Kutz, M. (Ed.), Standard Handbook of Biomedical Engineering & Design. McGraw-Hill, New York, pp. 35.17–35.31.

Reinkensmeyer, D.J., Emken, J.L., Cramer, S.C., 2004. Robotics, motor learning, and neurologic recovery. Annual Review of Biomedical Engineering 6, 497–525.

Reynolds, D.B., Repperger, D.W., Phillips, C.A., Bandry, G., 2003. Modeling the dynamic characteristics of pneumatic muscle. Annals of Biomedical Engineering 31, 310–317.

Sall, J., Lehman, A., Creighton, L., 2001. JMP Start Statistics A Guide to Statistics and Data Analysis Using JMP and JMP IN Software, Second ed. Duxbury, Pacific Grove, CA.

Scheidt, R.A., Reinkensmeyer, D.J., Conditt, M.A., Rymer, W.Z., Mussa-Ivaldi, F.A., 2000. Persistence of motor adaptation during constrained, multi-joint, arm movements. Journal of Neurophysiology 84, 853–862.

Shadmehr, R., Moussavi, Z.M.K., 2000. Spatial generalization from learning dynamics of reaching movements. Journal of Neuroscience 20, 7807–7815.

Shadmehr, R., Mussa-Ivaldi, F.A., 1994. Adaptive representation of dynamics during learning of a motor task. Journal of Neuroscience 14, 3208–3224.

Tondu, B., Lopez, P., 2000. Modeling and control of McKibben artificial muscle robot actuators. IEEE Control Systems Magazine 20, 15–38.

Wernig, A., Muller, S., Nanassy, A., Cagol, E., 1995. Laufband therapy based on 'rules of spinal locomotion' is effective in spinal cord injured persons. European Journal of Neuroscience 7, 823–829.

Winter, D.A., 1983. Energy generation and absorption at the ankle and knee during fast, natural, and slow cadences. Clinical Orthopaedics, 147–154.

Zatsiorsky, V.M., 2002. Kinetics of Human Motion. Human Kinetics, Champaign, IL.