# An Exoskeleton Using Controlled Energy Storage and Release to Aid Ankle Propulsion

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*Abstract* — **Symmetric ankle propulsion is the cornerstone of efficient human walking. The ankle plantar flexors provide the majority of the mechanical work for the step-to-step transition and much of this work is delivered via elastic recoil from the Achilles' tendon - making it highly efficient. Even though the plantar flexors play a central role in propulsion, body-weight support and swing initiation during walking, very few assistive devices have focused on aiding ankle plantarflexion. Our goal was to develop a portable ankle exoskeleton taking inspiration from the passive elastic mechanisms at play in the human triceps surae-Achilles' tendon complex during walking. The challenge was to use parallel springs to provide ankle joint mechanical assistance during stance phase but allow free ankle rotation during swing phase. To do this we developed a novel 'smart-clutch' that can engage and disengage a parallel spring based only on ankle kinematic state. The system is purely passive - containing no motors, electronics or external power supply. This 'energy-neutral' ankle exoskeleton could be used to restore symmetry and reduce metabolic energy expenditure of walking in populations with weak ankle plantar flexors (e.g. stroke, spinal cord injury, normal aging).**

**Keywords- human walking, ankle exoskeleton, plantar flexors, elastic energy storage and return, passive dynamics, 'energy-neutral', metabolic cost**

## I.   BACKGROUND

Coordinated ankle propulsion is a critical factor for efficient human walking. The ankle plantar flexors contribute the majority of the mechanical work done on the center-of-mass during push-off (Phase 4, 50-60% Stride) (Fig. 1). The efficiency of human gait is particularly impacted by the timing of the push-off and collision impulsive ground reaction forces [1]. For example, following stroke, propulsive impulses delivered by the ankle plantar flexors are often highly asymmetric. Push-off asymmetry results in slow walking speeds and increased metabolic energy consumption [3-5]. Due to the vital role of the ankle plantar flexors in shaping the normal mechanics and energetics of walking, it is essential to investigate ways in which we can improve gait impairments by focusing on aiding ankle joint push-off.

Portable wearable robotic devices hold considerable promise for restoring ankle function in populations with musculoskeletal or neurological impairments. In general,
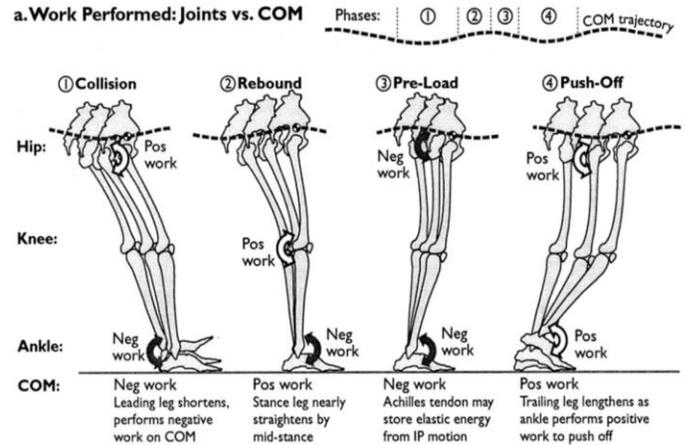


**Figure 1. Joint and Center of Mass Mechanics of Walking**
(a) Schematic detailing the phases of mechanical energy generation (Pos work) and absorption (Neg work) measured via inverse dynamics at the ankle, knee and hip joints during walking at an intermediate speed.
(b) Instantaneous work rate (i.e. mechanical power) (W/kg) over the stride cycle from heel strike (0%) to heel strike (100%) of the same leg. Note that during push-off (Phase 4, highlighted in red) the ankle plantar flexors generate the majority of the total positive work performed on the center-of-mass.
*Figure is adapted from* [1].
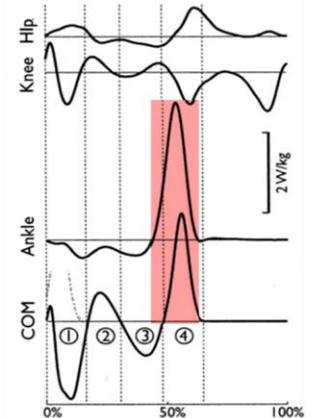
current devices fall into two distinct categories (1) fully-powered [6-9] and (2) purely passive [10]. Fully-powered devices employ motors under high gain force control that can mimic the normal torque output of the lower-limb joints over the full gait cycle. Some major downsides to this approach are that powerful motors are heavy, require bulky gears and mounting frames, and rely on finite power sources that must be donned by the user. The consequence of this added mass is a marked decrease in walking economy (i.e. no metabolic savings) during assisted locomotion with portable powered devices.

Purely passive devices (e.g. dynamic ankle-foot orthoses (DAFOs)) can store and release elastic energy in rigid, non-hinged frames to assist walking without assistance from motors. The main advantages of DAFOs are that they are lightweight, low cost and easy to maintain. Furthermore, recent work has shown that DAFOs can lead to small increases in both walking speed and economy post-stroke [11-13].

There are two key downsides to current DAFO designs. First, rigid, non-hinged DAFOs restrict full ankle joint range of motion, allowing only limited rotation in the sagittal plane. Second, and perhaps more crucial- current DAFOs do not allow free ankle rotation during swing, making it difficult to dorsiflex in preparation for heel strike. Inability to dorsiflex freely during swing could impose a significant metabolic penalty.

Key components from both the purely passive and fully powered wearable ankle devices can be combined in a 'hybrid' approach that offers optimized mobility assistance. For example, by actively controlling the passive elastic properties of an ankle exoskeleton it should be possible to produce the normal torque output of the ankle joint during walking with minimal actuation. This concept is analogous to the elastic 'catapult' mechanism observed in the human ankle during walking [2, 14]. In that case, the plantar flexor muscles (soleus, gastrocnemius) are activated nearly isometrically during stance to provide a rigid attachment for the Achilles' tendon to stretch against (Fig. 2). This 'muscle actuated latch' allows natural rotation of the center-of-mass over the ankle joint (i.e. inverted pendulum motion) during single limb support to transfer energy to the Achilles' tendon as it is stretched performing negative work (see Fig. 1, Phases 2-3). Then, at terminal stance the stored strain energy in the Achilles' is rapidly returned to the body, powering push-off (Fig. 1, Phase 4) [1, 2]. During swing phase plantar flexor muscles relax, and ankle dorsiflexors can reposition the foot for heel strike with no resistance from antagonists.

Evidence suggests that human walkers save considerable metabolic energy by exploiting the passive dynamic principle of elastic energy storage and return via the Achilles' tendon [15]. We hypothesize that a passive ankle exoskeleton using a parallel spring during the walking cycle is capable of recycling a significant portion of ankle joint mechanical work. A recent study indicates that when humans don tethered (i.e. non-portable), bilateral, lightweight, pneumatically powered ankle exoskeletons that replace *only* ~63% of the ankle muscle-tendon mechanical work during push-off, they can reduce metabolic energy consumption by 4% compared to walking without exoskeletons (with an added mass penalty of ~6%) [16]. Thus, supplying mechanical energy at a single joint (i.e. the ankle) during the key propulsive phase of walking (i.e. push-off) can have appreciable metabolic benefits. By taking a lightweight, portable approach it should be possible to reduce the metabolic cost of unimpaired walking by up to 18% [15]. In clinical populations (e.g. stroke) it should be possible to

reduce metabolic cost by up to 30%, bringing energy consumption into normal ranges on a mass specific basis.
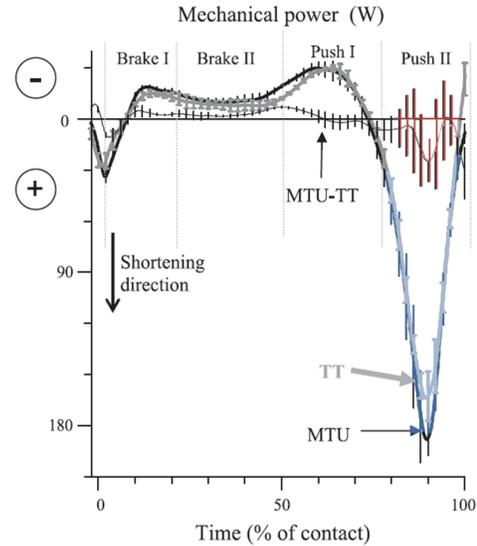


**Figure 2. Mechanics of Plantar Flexor Muscle-Tendon Interaction** Mechanical power (watts) produced by the soleus-Achilles' muscle-tendon unit (MTU), tendinous tissues (TT) and the remainder (MTU-TT), which could be attributed to soleus muscle fascicles plotted over the ground contact phase (% of contact) of walking. Note that during Push II (same as Fig. 1, Phase 4, ankle push-off) the soleus fascicles (in red) remain nearly isometric, and the elastic tissues (in blue) do most of the positive ankle joint work during push-off.  *Figure adapted from* [2].

II.  DESIGN

| Table 1. | Key Design Specifications |
|---|---|
| Linear Spring Stiffness, ($k_{eff}$) | 23.4 N/mm |
| Moment Arm | 126 mm |
| Max Torque | 109 Nm |
| Elastic Energy Stored in Spring | 20.7 J |

A. *Goals*

The main objective of our research is to build a passive elastic ankle exoskeleton that can reduce the metabolic cost of human walking. We took a biologically inspired design approach: exploiting the mechanism of controlled elastic energy storage and release observed in the human ankle during walking. By harnessing the passive dynamics inherent in human movement, our device will provide all the benefits of an actively powered system (e.g. [16]) but in a lightweight, portable framework (i.e. non-tethered) without motors, electronics or an external energy source.

B. *Key Design Objectives*

(1) To deliver torque to the ankle following a pattern similar to the normal joint moment during walking (Fig. 1, 3 and 4) in a portable, passive framework

(2) To recycle elastic energy during the stance phase while allowing free ankle rotation during swing phase (i.e. controlled energy storage and release).

## C. Methods

To generate torque in parallel with the ankle joint center and match the normal ankle joint moment we centered our design around a single commercially available linear tension spring (see Table 1 for specifications). We calculated the initial value of $k_{eff}$ (an upper limit) in order to achieve 100% of the peak ankle moment based on inverse dynamics data from human walking (Fig. 3). We will perform human walking trials to determine an optimal stiffness- the one that leads to the lowest metabolic energy cost.

In order to properly store and release the spring's energy at key intervals during the stride a mechanical control system was needed. We designed a system of springs, pins, and motion constraints to control the latch and release of a ratchet and pawl, engaging and disengaging the parallel springs during walking (Fig. 5, 6). This novel, adjustable 'smart-clutch' is advantageous because it uses the linear motion of the spring linkage, transmitted by changes in ankle joint angle, rather than electro-mechanical switching to set the timing of pawl latch and release (Fig. 7). We attached the 'smart-clutch' (mass=143g) and linear tension spring to a lightweight aluminum and carbon fiber ankle exoskeleton (mass=427g) (Fig. 5). We designed the exoskeleton to house the clutch at the back of the calf section of the shank upright. The clutch's Kevlar strands are rigidly connected to the linear tension spring, which is attached to the heel section of the exoskeleton in a position with a moment arm that generates similar torque to that of human walking.
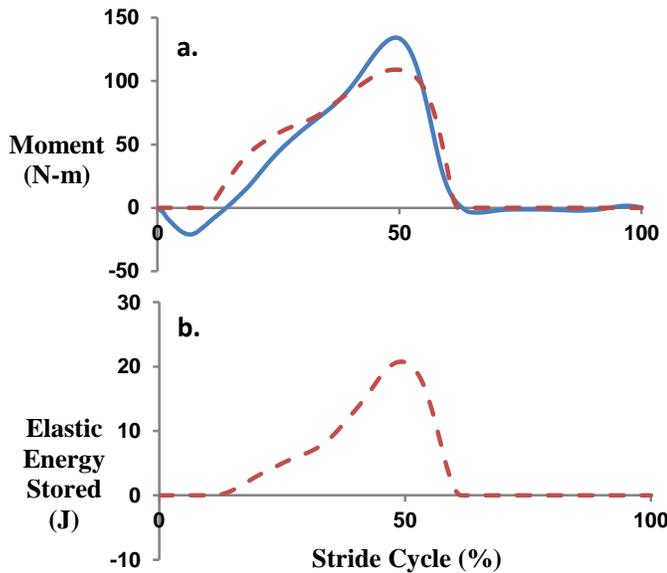


**Figure 3. Ankle Exoskeleton Mechanical Assistance During Walking**
(a) Calculated torque (N-m) contributed by the exoskeleton parallel spring (red/dashed) compared with the net muscle moment (N-m) measured via inverse dynamics at the ankle joint (blue) during normal human walking at 1.25 m/s. Plantar flexor moment is positive. (b) Elastic energy stored (J) in the exoskeleton spring from foot-flat (~20% stride cycle) to max dorsiflexion (~50% stride cycle) due to rotation of the center-of-mass about the ankle joint. For these design specifications (see Table 1.) the elastic energy stored in the spring is returned rapidly during push-off (~50-60% stride cycle) and provides *all* of the ankle positive work for propulsion. For both panels, data are plotted for a single stride from heel-strike (0%) to heel-strike (100%) of the same leg.

## III. CLUTCH DESIGN IN DEPTH

The central component to our design is a clutching mechanism that employs the principle of controlled energy storage and release (CESR) [17]. A 'zoomed-in' look at the biomechanics of muscle-tendon interaction of the ankle plantar flexors motivated our bio-inspired approach to the clutch design. Ultrasound imaging studies provide an important window into the mechanisms behind mechanical work production of the triceps surae muscles (soleus and gastrocnemius) and the in-series Achilles' tendon [2]. These studies reveal that most or even all of the positive work produced during ankle push-off may come from the recoil of previously stretched elastic tissues (Fig. 2). In addition, the soleus and gastrocnemius muscle fascicles produce high forces during *nearly isometric* contractions. The strut-like behavior of an isometric muscle is mechanically similar to a static motion constraint that can be turned on when the muscle is activated and turned off when the muscles is relaxed (i.e. a clutch transmission) [18]. The series elastic tissues (Achilles' tendon and aponeurosis) can be easily substituted by mechanical springs or synthetic elastic elements [19, 20].

Taking into account the normal onset timing of muscle activation of the soleus and gastrocnemius muscles we determined the key points in the walking stride to engage (~10% stride cycle) and disengage (~60% of the stride) the clutch (Fig. 7). The timing of disengagement is important because it allows free rotation of the ankle joint during swing- a novel aspect of our design compared to traditional DAFOs.
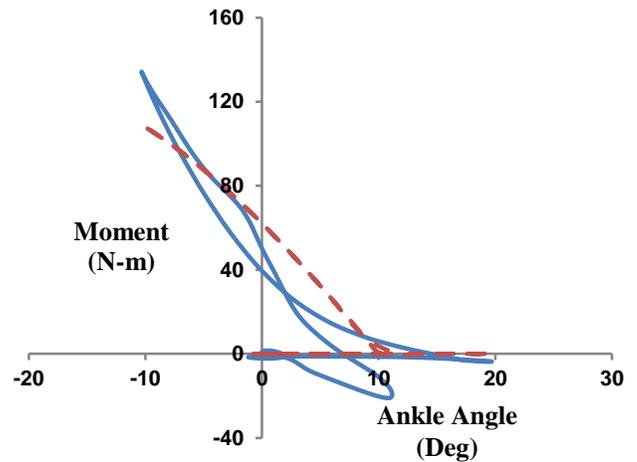


**Figure 4. Human Ankle Work-Loops During Assisted Walking**
Calculated exoskeleton torque (N-m) (red/dashed curve) and ankle joint moment (N-m) (blue curve) versus ankle joint angle (deg) plotted over a full walking stride. The small inscribed area within the ankle joint work-loop (blue) indicates pseudo-spring-like behavior of the joint. To emulate these mechanics with our elastic exoskeleton we chose a spring and moment arm combination to generate torsional stiffness as close as possible to normal walking at 1.25 m/s. For both angles and moments + indicates plantarflexion/flexor.
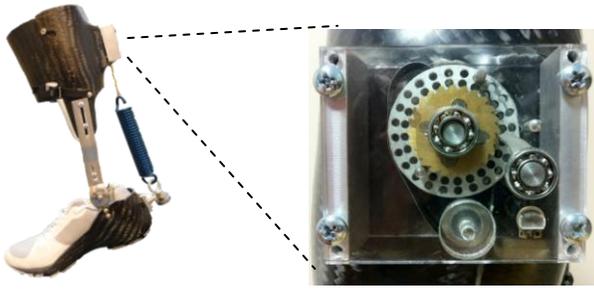
**Figure 5. 'Smart-Clutch' for Controlled Energy Storage and Release**
Prototype clutch with plexi-glass cover (right) and exoskeleton with the clutch coupled to a series spring providing parallel ankle stiffness (left).

As shown in Fig. 7 the clutch starts by allowing downward movement of the linkage (Kevlar strands plus linear spring) until heel strike (in orange). At maximum dorsiflexion, a timing pin engages the ratchet and pawl mechanism which restricts further downward movement of the linkage. Then the reaction force from a constant tension spring in the clutch takes up slack in the system as the foot plantar flexes until foot-flat position (in purple). Following foot-flat, as the ankle begins to dorsiflex into mid stance, the clutch locks and the linkage transmits force to stretch the linear spring, storing energy from the body's center-of-mass (in green). During push-off, all of energy that was stored in the linear spring is returned at the ankle joint to perform positive
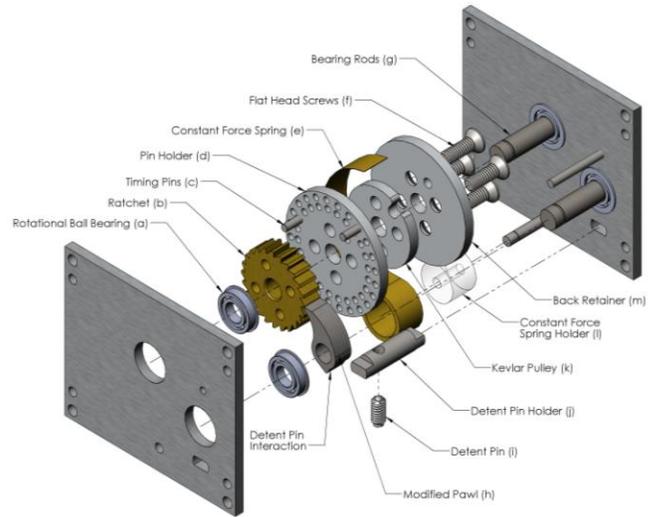


**Figure 6. Exploded View of 'Smart-Clutch' Components**
Important features to note include: (b) ratchet (c) timing pins (h) pawl. Ankle joint movements are transmitted to rotation of the ratchet through the Kevlar and linear spring linkage over the stride. Timing pins are set to initiate and terminate engagement of the pawl at key points during

mechanical work, propelling the body forward (in dark blue). Then, a second timing pin disengages the ratchet and pawl mechanism (in gray) at maximum plantar flexion and frees up the ankle to dorsiflex during swing, resetting the cycle.
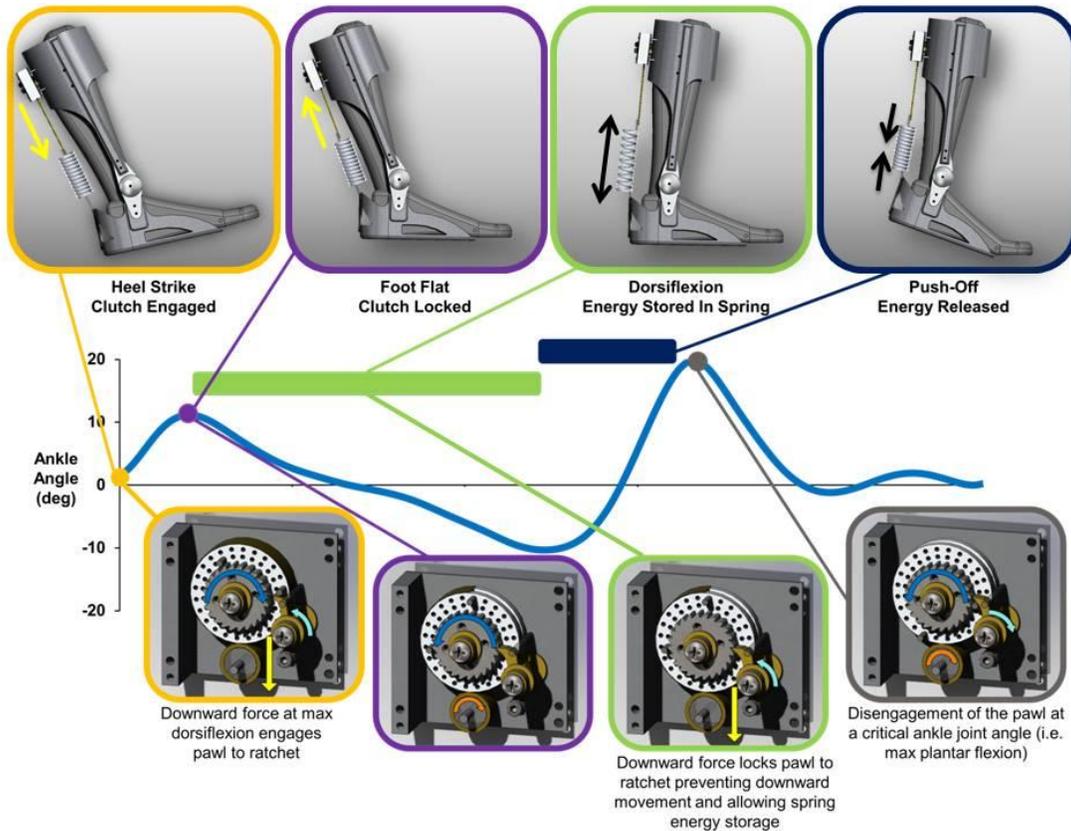


**Figure 7. Passive Elastic Ankle Exoskeleton Function**
Schematic highlighting the key events of the 'smart-clutch' function over the walking stride. Normal ankle joint angle profile from walking at 1.25 m/s is shown in blue (+ is plantarflexion) to demonstrate the direct coupling between ankle kinematics and the clutch timing.

## IV. CONCLUSIONS

Our biologically-inspired ankle exoskeleton design emulates the elastic energy storage and return cycle observed in the human triceps surae-Achilles' muscle-tendon unit during human walking. The current prototype responds directly to mechanical feedback from the ankle angle in order to engage and disengage a parallel elastic spring. The novelty of our design is that it is able to switch between an energy storage mode during stance and a free rotation mode during swing in a purely-passive (no motors or electronic control) package. These design features eliminate the need for an external power source and make it portable and lightweight. Even with its streamlined design, bench top tests of the prototype show that it easily supports the full body weight of a typical user (body mass = 75kg) with a factor of safety of 2.5.

One limitation to our current prototype is that the phases of energy storage and release, and free rotation are set for a particular gait (i.e. steady walking at a given pre-selected speed) and must be manually adjusted for changes in gait. We have begun to develop a next generation prototype that accounts for this limitation by automatically adjusting for changes in gait using mechanical sensing and feedback, without the use of electronics.

Next steps include: (1) testing the 'smart-clutch' on the bench top to verify robust behavior over many cycles (2) performing human walking tests (in both impaired and unimpaired subjects) to determine whether the device can reduce metabolic energy expenditure at different speeds and spring stiffnesses (3) continuing to develop the next generation prototype with the capability to adjust the timing of engagement of the clutch 'on the fly'. Adjustable clutch timing will be important for performance during walking tasks with changing mechanical demands (i.e. increases in speed).

The benefits of a functional passive elastic exoskeleton focused on aiding ankle 'push-off' would be widespread. This device could be used to restore normal walking in people who have gait impairments following stroke, spinal cord injury, cerebral palsy, Achilles' tendon rupture or even arthritis. In addition, this device could be used to aid soldiers in the field or for strenuous recreational hiking. Finally, establishing the limits of what is possible with a purely passive exoskeleton will be beneficial for those designing actuated systems for applications in lower-limb prosthetics and autonomous legged robotics.

## ACKNOWLEDGMENT

## REFERENCES

[1] A. D. Kuo, *et al.*, "Energetic consequences of walking like an inverted pendulum: step-to-step transitions," *Exerc Sport Sci Rev,* vol. 33, pp. 88-97, Apr 2005.

[2] M. Ishikawa, *et al.*, "Muscle-tendon interaction and elastic energy usage in human walking," *J Appl Physiol,* vol. 99, pp. 603-8, Aug 2005.

[3] G. Chen and C. Patten, "Joint moment work during the stance-to-swing transition in hemiparetic subjects," *J Biomech,* vol. 41, pp. 877-83, 2008.

[4] S. Nadeau, *et al.*, "Plantarflexor weakness as a limiting factor of gait speed in stroke subjects and the compensating role of hip flexors," *Clinical Biomechanics,* vol. 14, pp. 125-35, Feb 1999.

[5] C. Detrembleur, *et al.*, "Energy cost, mechanical work, and efficiency of hemiparetic walking," *Gait Posture,* vol. 18, pp. 47-55, Oct 2003.

[6] H. Herr, "Exoskeletons and orthoses: classification, design challenges and future directions," *J Neuroeng Rehabil,* vol. 6, p. 21, 2009.

[7] A. M. Dollar and H. Herr, "Lower extremity exoskeletons and active orthoses: Challenges and state-of-the-art," *Ieee Transactions on Robotics,* vol. 24, pp. 144-158, Feb 2008.

[8] J. Hitt, *et al.*, "Dynamically controlled ankle-foot orthosis (DCO) with regenerative kinetics: Incrementally attaining user portability," *Proceedings of the 2007 Ieee International Conference on Robotics and Automation, Vols 1-10,* pp. 1541-1546, 2007.

[9] H. Kazerooni, *et al.*, "That which does not stabilize, will only make us stronger," vol. 26, pp. 75-89, Jan 2007.

[10] M. C. Faustini, *et al.*, "Manufacture of passive dynamic ankle-foot orthoses using selective laser sintering," *Ieee Transactions on Biomedical Engineering,* vol. 55, pp. 784-790, Feb 2008.

[11] A. Danielsson and K. S. Sunnerhagen, "Energy expenditure in stroke subjects walking with a carbon composite ankle foot orthosis," *Journal of Rehabilitation Medicine,* vol. 36, pp. 165-168, Jul 2004.

[12] K. J. Nolan, *et al.*, "Evaluation of a dynamic ankle foot orthosis in hemiplegic gait: A case report," *NeuroRehabilitation,* vol. 27, pp. 343-50, 2010.

[13] M. Franceschini, *et al.*, "Effects of an ankle-foot orthosis on spatiotemporal parameters and energy cost of hemiparetic gait," *Clinical Rehabilitation,* vol. 17, pp. 368-372, Jul 2003.

[14] T. Fukunaga, *et al.*, "In vivo behaviour of human muscle tendon during walking," *Proceedings of the Royal Society of London: Biological Sciences,* vol. 268, pp. 229-33, Feb 7 2001.

[15] G. S. Sawicki, *et al.*, "It pays to have a spring in your step," *Exerc Sport Sci Rev,* vol. 37, pp. 130-8, Jul 2009.

[16] G. S. Sawicki and D. P. Ferris, "Powered ankle exoskeletons reveal the metabolic cost of plantar flexor mechanical work during walking with longer steps at constant step frequency," *J Exp Biol,* vol. 212, pp. 21-31, Jan 2009.

[17] S. H. Collins and A. D. Kuo, "Recycling energy to restore impaired ankle function during human walking," *PLoS One,* vol. 5, p. e9307, 2010.

[18] K. Endo and H. Herr, "A Model of Muscle-Tendon Function in Human Walking," *Icra: 2009 Ieee International Conference on Robotics and Automation, Vols 1-7,* pp. 2790-2796, 2009.

[19] K. Endo, *et al.*, "An artificial gastrocnemius for a transtibial prosthesis," *Conf Proc IEEE Eng Med Biol Soc,* vol. 2009, pp. 5034-7, 2009.

[20] A. M. Dollar and H. Herr, "Design of a Quasi-Passive Knee Exoskeleton to Assist Running," *2008 Ieee/Rsj International Conference on Robots and Intelligent Systems, Vols 1-3, Conference Proceedings,* pp. 747-754, 2008.