The Biomechanics and Energetics of Human Running Using an Elastic Knee Exoskeleton

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Abstract—While the effects of series compliance on running biomechanics are well documented, the effects of parallel compliance are known only for the simpler case of hopping. As many practical exoskeletal and orthotic designs act in parallel with the leg, it is desirable to understand the effects of such an intervention. Spring-like forces offer a natural choice of perturbation for running, as they are both biologically motivated and energetically inexpensive to implement. To this end, we investigate the hypothesis that the addition of an external elastic element at the knee during the stance phase of running results in a reduction in knee extensor activation so that total joint quasi-stiffness is maintained. An exoskeletal knee brace consisting of an elastic element engaged by a clutch is used to provide this stance phase extensor torque. Motion capture of five subjects is used to investigate the consequences of running with this device. No significant change in leg stiffness or total knee stiffness is observed due to the activation of the clutched parallel knee spring. However, this pilot data suggests differing responses between casual runners and competitive long-distance runners, whose total knee torque is increased by the device. Such a relationship between past training and effective utilization of an external force is suggestive of limitations on the applicability of assistive devices.

I. INTRODUCTION

It is a long standing result in biomechanics that running most resembles a mass bouncing on a linear spring [2]–[5]. Previous work [6] has demonstrated the efficacy of a parallel spring spanning the leg during hopping, but such investigation is more challenging in running, due in large part to the difficulty of applying such an external force while still permitting knee flexion during swing. A custom clutch-spring exoskeleton spanning the knee allows such an intervention during stance phase only.

Running is typically modeled as a mass rebounding off of a constant linear stiffness spring [2], [3]. This stiffness may be purely vertical or may be fixed to the biological leg, rotating in the in the sagittal plane and changing its angle relative to the ground during stance. Following McMahon & Cheng [3], this spring may be considered as exerting a purely vertical force with a stiffness given by

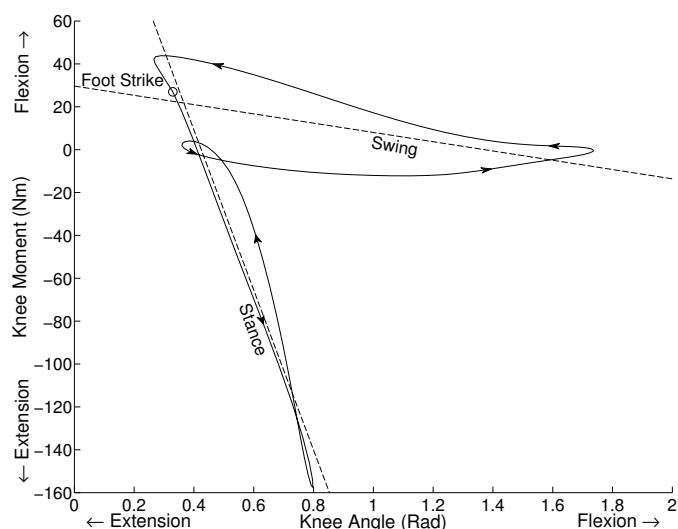$$k_{vert} = \frac{F_{z,peak}}{\Delta y} \qquad (1)$$

Fig. 1.   Torque versus angle of the knee during running. Dashed lines indicate least squares linear fits during stance and swing showing two stiffness behaviors during these phases of the running gait. Plot based on data from [1].

where $F_{z,peak}$ is the maximum vertical component of the ground reaction force and Δy is the vertical displacement of the center of mass. Due to the angle subtended by the leg, however, an effective spring tracking the rotation of the leg is compressed from its rest length L_0 by ΔL much larger than Δy, so that

$$k_{leg} = \frac{F_{z,peak}}{\Delta L} \qquad (2)$$

Assuming symmetry of absorptive and generative stance with θ representing half the total subtended angle, it may be shown geometrically that ΔL is given by

$$\Delta L = \Delta y + L_0 \left(1 - \cos\theta\right) \qquad (3)$$

Moreover, it can be shown [3] that the effective θ is a function of forward velocity u and ground contact time t_c:

$$\theta = \sin^{-1}\left(\frac{ut_c}{2L_0}\right) \qquad (4)$$

This k_{leg} is used to characterize running gaits and must, by definition, be smaller than k_{vert}. For typical running speeds (3-5 m/s), k_{leg} is on the order of 10kN/m and varies relatively little with speed [3], [4]. However, leg stiffness does increase, for instance, if stride frequency is deliberately increased [7].

Critically, this spring characterizes the behavior only of the stance leg. During swing, the effective stiffness decreases dramatically, as the knee flexes, lowering the moment of inertia of the leg about the hip. This two-stiffness model applies not only to the leg on the whole, but to individual joints. As shown in Figure 1, the knee is similarly characterized by a very high stiffness during stance, but nearly zero stiffness during swing.

These stiffnesses likely derive from energy storage and transfer in tendons and ligaments [8]–[10]. Such a passive mechanism accounts in part for the efficiency of human running, though it is important to note that even ideal energy storage of this kind is not without cost as the series muscle must exert an opposing force to enable tendon stretch. Even a nearly isometric contraction, yielding little or no net mechanical work incurs a significant metabolic cost [9], [11].

Recently, exoskeletons have sought to emulate this passive elastic architecture for augmentative purposes. One may hypothesize that if the role of tendon and other elastic tissue could be fulfilled externally, series muscle activation could be reduced or eliminated while total kinetics, with the exoskeletal contribution included, are preserved. In fact, there is evidence that total leg stiffness, including external contributions, is maintained in bouncing gaits. If a series elasticity, namely a compliant ground surface, is introduced, total k_{vert} including the series compliant surface is maintained [12], [13], even when doing so requires increasing biological leg stiffness by up to 68%. This adaptation is extremely fast, occurring within the very first step after transition to the compliant surface [14].

One exoskeleton of this type, designed by Wiggin, et al. [15], places a clutch and spring in parallel with the ankle in order to assist powered plantar flexion in walking. Another, by Cherry, et al. [16] couples the knee via bowden cable to a clutch and spring worn on a backpack, which provides an extensor moment during running stance. Cherry's device additionally introduces a purely passive spring spanning the ankle.

Here, a novel mechanical clutch and leaf spring system is used to replicate the two-stiffness behavior external of the knee, providing a stiffness during stance, but not during swing. In this investigation, using such a device, the effects of an external joint stiffness on biological joint and leg stiffnesses was assessed experimentally.

II. METHODS

The exoskeleton used in this study consists of a composite leaf spring in parallel with the leg, articulated at its midpoint with a clutch so that the spring may collapse during swing phase, providing no elastic force, but lock prior to stance. This architecture, depicted in Figure 2, represents a modification to the hopping exoskeleton demonstrated by Grabowski [6], though here it is used to span only the knee, as shown in Figure 3, rather than the entire leg.
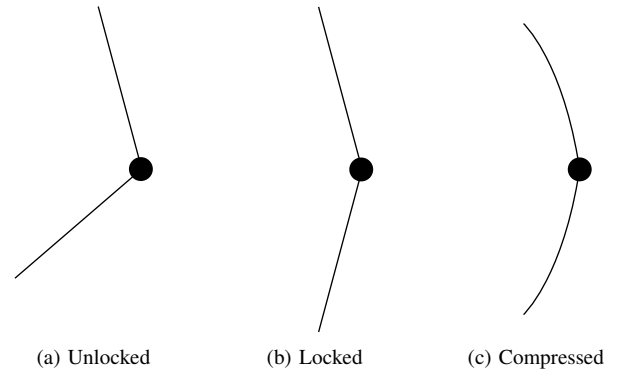

(a) Unlocked   (b) Locked   (c) Compressed

Fig. 2. Conceptual behavior of a collapsible bow spring with exoskeletal knee joint, as used to augment running

Fig. 3. Clutch-spring knee exoskeleton, as worn here.

A. Exoskeleton

A novel clutch design is necessary to simultaneously achieve high (up to 190Nm) holding torque and high (1.8°) locking resolution. The exoskeletal knee consists of an interference clutch (chosen for its high ratio of holding torque to mass) in series with a planetary gearbox which decreases load on the clutch and increases effective resolution, compensating for the discretized engagement of the interference clutch. The complete device is constructed so that the distal bow spring attachment serves as input to the ring of a planetary system whose planet carrier is fixed to the proximal assembly. The output of the planetary system (the sun) is coupled through a toothed clutch to the proximal assembly, effectively locking the joint when the clutch is engaged by activation of a solenoid.

The harness used to attach the exoskeleton to the thigh and calf regions is based on an elongated polycentric carbon fiber knee brace, as shown in Figure 3. On either side of the knee, the exoskeleton applies load to the leg through rubber-lined carbon fiber cuffs. The cuffs have been designed to avoid muscle bellies, including the hamstrings and gastrocnemius that are active in running. The upper and lowermost shank portions of the harness are composed of rigid carbon fiber strips. Four independently adjustable velcro straps provide additional support to both anterior and posterior portions of the brace. The lateral attachment of the leaf springs allow the exoskeleton to follow the form of the leg relatively closely, but results in a larger than normal torque in the coronal plane during stance. To minimize this effect, the lateral displacement has been reduced as much as possible.

An onboard microcontroller infers phase of gait cycle using an encoder on the exoskeletal joint and a three degree-of-freedom inertial unit to advance a state machine. The control strategy is such that the clutch is fully locked at the peak knee extension which immediately precedes heel strike and remains locked until toe-off.

B. Protocol

The proposed exoskeleton provides an elastic element in parallel with each knee during stance phase, but unfortunately a practical device also influences the body in several other ways due to its mass and means of attachment. The added mass of the exoskeleton has a gravitational effect on hip extensors and knee flexors, as they must lift the exoskeletal mass during early swing. In addition, the added mass also has an inertial effect on hip flexors, as they must accelerate the mass during swing. Finally, attachment to the body is difficult to accomplish without some constriction, which limits range of motion and causes discomfort. In order to isolate the effect of elasticity, experiments were conducted in three conditions: control, in which subjects ran in self-selected footwear with no experimental apparatus other than those required for instrumentation; inactive, in which subjects wore the investigational knee braces with the power off, contributing zero stiffness but offering the same secondary effects associated with mass and restricted movement; and active, in which subjects wore the investigational knee braces with the power on, contributing a non-zero parallel stiffness during stance phase. In all trials, the exoskeleton was worn bilaterally.

Human testing was conducted in accordance with MIT COUHES protocol number 0801002566 and UNC IRB protocol number 10-0691. Pilot trials and device testing were conducted in the MIT Biomechatronics Group. Experimental trials were conducted in the NCSU PoWeR Laboratory.

Six male subjects (average mass 69 ± 8kg and height $181\pm 8cm$), described in Table II, were recruited from a pool of healthy recreational runners having leg length (>90cm) and circumference (45-55cm at the thigh, 20-30cm at the shin) consistent with the investigational knee brace.

Each subject ran with the device active for a training session of at least thirty minutes on a day prior to instrumented trials. Subjects trained first on open ground then on a treadmill wearing a fall prevention harness (Bioness, Valencia, CA, USA). During this training session, subjects with a gait insufficiently wide to prevent collision between the braces, or with stance knee extension insufficient to ensure disengagement of the clutch, were disqualified on the basis of safety.

During the experimental session, a nominal 0.9 Nm/^{o} elastic element was used. This relatively small stiffness proved necessary due to the effects of series compliance in the harness and the tendency of the biological knee to resist a stiffer exoskeleton by shifting anteriorly in the brace.

At the start of the experimental session, each subject's self-selected step frequency was measured while running on the treadmill at 3.5m/s without the investigational knee brace. The time necessary to complete 30 strides was measured by stopwatch after approximately one minute of running. This cadence (166 ± 9 steps/s) was enforced by metronome for all subsequent trials.

After being instrumented for electromyography and motion capture, subjects then ran on the instrumented treadmill at 3.5m/s in the control, inactive, and active conditions. Trial order was randomized, excepting that inactive and active conditions were required to be adjacent, so as to require only a single fitting of the investigational device in each session. Each running trial was seven minutes in length, with an intervening rest period of at least as long. Resting metabolism was also measured for five minutes at both the start and end of the experimental session. Sessions lasted approximately three hours, including 21 minutes of treadmill running.

C. Instrumentation and Processing

During the experimental session, each subject was instrumented for joint kinematics and kinetics, electromyography, and metabolic demand.

Subject motion was recorded using an 8 camera passive marker motion capture system (VICON, Oxford, UK). Reflective markers were adhered to subjects using a modified Cleveland Clinic marker set for the pelvis and right leg (Left and right ASIS and Trochanter, three marker pelvis cluster, four marker thigh cluster, medial and lateral epicondyle, four marker shin cluster, medial and lateral malleolus, calcaneus,

Fig. 4. Right leg instrumented for motion capture. Black tape covers all reflective surfaces.

foot, fifth metatarsal). For inactive and active trials, the termination points of the exoskeletal spring were also marked. The marker set for the right leg is shown in Figure 4. Motion data were recorded at 120Hz and low pass filtered using a 2nd order Butterworth filter with a 10Hz cutoff. Ground reaction forces were recorded at 960Hz from a dual belt instrumented treadmill (BERTEC, Columbus, OH, USA) and low pass filtered using a 2nd order Butterworth filter with a 35Hz cutoff. Following calibration using a static standing trial, Visual3D (C-Motion Inc, Germantown, MD, USA) modeling software was used to reconstruct joint kinematics and kinetics and center of mass trajectories, assuming right-left leg symmetry.

Fifty steps from each trial were analyzed to determine average leg and joint stiffness. Due to technical difficulties associated with loss or migration of motion capture markers and the appearance of false markers due to reflectivity of the exoskeleton, some motion capture recordings proved unusable. Consequently, the exact timing of the steps used varies between subjects and it was not possible to analyze fifty steps for all trials. In general, the earliest available reconstructions a minimum of one minute into the trial were used, to minimize effects of fatigue.

k_{vert} and k_{leg} were calculated for each step using Equation 1 and Equation 2 with center of mass displacements determined by Visual3D through integration of ground reaction forces as in [17]. Unlike the effective leg spring, the knee and ankle experience different stiffnesses in absorptive (early) stance and generative (late) stance. Stiffnesses of these joints were estimated individually for the two phases using

$$\kappa_{joint,abs} = \frac{M_{joint,peak} - M_{joint,HS}}{\theta_{joint,peak} - \theta_{joint,HS}} \qquad (5)$$

$$\kappa_{joint,gen} = \frac{M_{joint,peak} - M_{joint,TO}}{\theta_{joint,peak} - \theta_{joint,TO}} \qquad (6)$$

where $peak$ represents the instant of peak torque in the joint and HS and TO represent heel-strike and toe-off, respectively.

Metabolic demand was measured noninvasively using a mobile cardiopulmonary exercise test system (VIASYS Healthcare, Yorba Linda, CA, USA), which measures rates of oxygen consumption and carbon dioxide production through a face mask. Once sub-maximal steady state metabolism was achieved, total metabolic power was deduced from linear expressions of the form

$$P = K_{O_2} V'_{O_2} + K_{CO_2} V'_{CO_2} \qquad (7)$$

where V'_{O_2} and V'_{CO_2} represent average rates of oxygen inhalation and carbon dioxide exhalation and K_{O_2} and K_{CO_2} are constants which have been well documented. Brockway's [18] values $K_O = 16.58kW/L$ and $K_{CO_2} = 4.51kW/L$ were used. Average rates were calculated over a two-minute window during steady state metabolism from 4:00 to 6:00 within each seven minute trial. In addition to the running conditions, resting metabolic power was also measured with the subject standing for five minutes.

Such measures of metabolic power are only valid if the contributions of anaerobic metabolism are small. This was assured by monitoring the ratio of volume of carbon dioxide exhaled to oxygen inhaled, known as the respiratory exchange ratio. Oxidative metabolism was presumed to dominate while this ratio was below 1.1.

Though the data are not discussed here, subjects were also instrumented with surface electromyography (EMG) electrodes on eight muscles of the right leg. Wires were taped to skin and routed an amplifier (Biomectrics Ltd, Ladysmith, VA, USA) clipped to the chest harness containing the cardiopulmonary test system. A reference electrode was attached to the wrist.

More details of the instrumentation used here can be found in [1], in which identical instrumentation and signal processing was used, with the omission of electromyography.

Repeated measures ANOVA was used to determine the significance of apparent differences between the control, inactive, and active conditions. For each measurement found to vary among the three groups, a post-hoc two-sided paired t-test was conducted using Šidák correction to compare the control and inactive conditions and inactive and active conditions.

III. RESULTS

Subjects S1, S2, S3, S4, and S5 exhibited similar gross kinematics in all three conditions. S6's mechanics are omitted, as he was visibly fatigued and failed to complete either the active or inactive trials. For each of the three conditions, stride averaged knee and leg stiffnesses calculated for each of subjects are presented in Table III and Table IV. Metabolic demand is presented in Table V.

Repeated measures ANOVA shows that vertical leg stiffness k_{vert} ($P = 0.01$) and leg stiffness k_{leg} ($P = 0.02$) vary among the conditions, with post-hoc paired t-testing revealing that the changes observed in k_{vert} and k_{leg} between the control and inactive conditions is significant ($P < 0.01$ in both cases), but that no significant difference exists between the inactive and active conditions. This evidence that increased mass at the knee increases leg stiffness is interesting, particularly in

TABLE III.   NORMALIZED KNEE STIFFNESS[1] $(Nm/kg/^o)$

	Control Biological	Inactive Biological	Active Total	Active Exo	Active Biological
S1	0.121±0.004	0.106±0.003	0.127±0.007	0.017±0.001	0.111±0.007
S2	0.114±0.003	0.113±0.003	0.115±0.002	0.025±0.001	0.090±0.002
S3	0.139±0.003	0.107±0.002	0.131±0.003	0.016±0.000	0.115±0.004
S4	0.089±0.002	0.116±0.002	0.111±0.002	0.011±0.000	0.101±0.002
S5	0.095±0.002	0.089±0.002	0.062±0.003	0.027±0.001	0.035±0.003
	0.112±0.020	0.106±0.010	0.110±0.028	0.019±0.007	0.090±0.032

(a) Absorptive Stance Knee Stiffness $\kappa_{knee,abs}$

	Control Biological	Inactive Biological	Active Total	Active Exo	Active Biological
S1	0.081±0.001	0.065±0.002	0.081±0.003	0.013±0.001	0.068±0.003
S2	0.102±0.001	0.084±0.002	0.087±0.001	0.019±0.000	0.068±0.001
S3	0.144±0.003	0.090±0.001	0.105±0.001	0.016±0.000	0.089±0.001
S4	0.091±0.001	0.088±0.001	0.089±0.001	0.010±0.000	0.079±0.001
S5	0.078±0.001	0.060±0.001	0.053±0.001	0.022±0.000	0.031±0.001
	0.099±0.027	0.077±0.014	0.083±0.019	0.016±0.005	0.067±0.022

(b) Generative Stance Knee Stiffness $\kappa_{knee,gen}$

[1]Uncertainties for individual subjects reflect the standard error of the mean over strides, while uncertainty for the mean reflects standard deviation associated with subject-to-subject variation

TABLE IV.   NORMALIZED LEG STIFFNESS[1] (Nm/kg)

	Control	Inactive	Active
S1	695±22	822±30	794±57
S2	609±15	690±19	709±16
S3	734±14	814±18	954±29
S4	500±8	593±18	530±7
S5	625±18	753±49	690±39
	633±90	734±95	735±155

(a) Vertical Leg Stiffness k_{vert}

	Control	Inactive	Active
S1	196±3	205±3	218±8
S2	191±2	198±2	204±1
S3	241±2	249±2	280±3
S4	153±1	165±2	155±1
S5	166±2	170±2	205±3
	189±34	197±34	212±45

(b) Leg Stiffness k_{leg}

[1]Uncertainties for individual subjects reflect the standard error of the mean over strides, while uncertainty for the mean reflects standard deviation associated with subject-to-subject variation

TABLE V.   NORMALIZED METABOLIC DEMAND[1] (W/kg)

	Resting	Control	Inactive	Active
S1	1.7±0.2	15.7±0.1	21.2±0.1	20.7±0.1
S2	1.3±0.2	17.2±0.2	19.6±0.3	19.7±0.3
S3	1.1±0.1	16.6±0.1	20.6±0.0	20.3±0.1
S4	1.4±0.1	16.6±0.1	19.6±0.1	20.4±0.1
S5[2]	1.2±0.1	17.2±0.3	16.9±0.1	-
	1.3±0.2	16.3±0.6	20.6±0.5	20.4±0.2

[1]Uncertainties for individual subjects reflect the standard error of the mean over breaths, while uncertainty for the mean reflects standard deviation associated with subject-to-subject variation
[2]Due to unavailability of metabolic data for subject S5 in the active condition, this subject is omitted from the means displayed

light of He's [19] finding that leg stiffness does not vary when gravity is reduced. If stiffness is normalized by total mass, only the increase in vertical stiffness is found to be significant, indicating that this term not only increases with mass but increases disproportionately.

ANOVA finds variation in total and biological knee stiffness in the generative phase ($P = 0.04$ and $P < 0.01$, respectively), but not in the absorptive phase. A post-hoc paired t-test, however, finds no significant change in knee stiffness between the control and inactive or inactive and active trials.

Finally, ANOVA finds significant variation in metabolic demand among the three conditions. In post-hoc testing, a suggestive difference exists between the control and inactive conditions ($P = 0.04$, not quite significant at the 5% level with the Šidák correction). This is misleading, however, as the respiratory exchange ratio is notably higher for trials in the inactive and active condition than for trials in the control condition. Though always below 1.1, this shift in respiratory exchange ratio implies that some anaerobic contribution is present when the brace is worn, making comparisons between the control and inactive case tenuous.

This pilot data provides no evidence against the null hypotheses that leg stiffness and knee stiffness are each unchanged by the presence of an external parallel spring at the knee. However, closer examination of Table III suggests that the population may be divided into two groups according to level of training. Only competitive marathoners S1 and S3 appear to exhibit increased total knee stiffness (in both absorption and generation) in the active condition relative to the inactive condition. Indeed, they appear to maintain their biological knee stiffness. While statistics for such a small sample must be approached very cautiously, a two sided paired t-test suggests increased total knee stiffness in both absorption and generation in marathoners ($P = 0.04$ and $P = 0.02$) with no corresponding effect in the remaining recreational runners ($P = 0.35$ and $P = 0.78$). Marathoners S1 and S3 also exhibit small (2%) reductions in metabolic demand above resting while S2 and S4 do not, though this effect is not

statistically significant. Verifying these apparent differences in stiffness and metabolic demand based on runner training would require subsequent investigation with larger samples of distance and casual runners, however.

IV. DISCUSSION

In the small group tested, there is no significant evidence of a change in biological leg stiffness in response to the addition of the parallel knee spring during running stance. There is some suggestion, however, of a correlation between running experience and knee stiffness in the presence of the intervention, such that casual runners regulate total knee stiffness while more trained distance runners allow total knee stiffness to increase. It is plausible that plasticity varies with level of training, so that recreational runners can reduce biological knee stiffness to preserve total stiffness in the presence of the external spring, while trained marathon runners cannot. It has been suggested that running dynamics arise predominantly from tuned morphology, with leg stiffness representing largely passive tendon stiffness [9], [10]. If this is the case, marathoners may have muscle fascicles optimized for low-speed, largely isometric contractions and therefore be physiologically less capable of modulating knee stiffness downward through spring-like muscle activation. Alternatively, it is possible that trained runners and recreational runners optimize different parameters while running. Recreational runners may optimize their gait for load reduction in joints or minimal center of mass displacement, while marathon runners may instead optimize their gait for metabolic demand.

This possible division in the population, only hinted at by this pilot data, would affect the applicability of interventions of this kind and therefore warrants further study. For wearers whose biological knee stiffness is down regulated in response to external stiffness, applications include orthoses that reduce pain in damaged joints by lowering loads, thereby restoring an active lifestyle to sufferers of chronic pain. Alternatively, for wearers whose total knee stiffness increases with the intervention, such a device could be used to supplement weak knee extensors. The relatively unencumbering design of this exoskeleton makes it particularly well suited to these and other quality of life applications which involve frequent wear. Though not investigated here, the device also has potential applicability in reducing metabolic demand while running under load, a challenge faced by soldiers and firefighters, for example.

Surprisingly, there is statistically significant evidence linking the addition of mass to the leg to an increase in leg stiffness and vertical leg stiffness. This effect may not entirely be due to the mass of the wearable knee brace, but may in part be due to the constraints it imposes on the wearer. Whatever the cause, this effect will be present in devices with this style of attachment and therefore should be considered when designing for augmentation or rehabilitation of running.

Biomechanically and physiologically, future work will likely focus on two areas: investigating the potential influence of past running training, and determining the functional dependence of metabolic demand and joint stiffnesses on the externally applied stiffness. Mechatronically, a critical area for future investigation is exoskeletal attachment so as to achieve efficient torque and power transfer from the exoskeletal device to the body, enabling investigation of higher external stiffnesses than could be used here.

REFERENCES

[1] D. J. Farris and G. S. Sawicki, "The mechanics and energetics of human walking and running: A joint level perspective," *Journal of the Royal Society Interface*, vol. 9, no. 66, pp. 110–118, 2011.

[2] R. Blickhan, "The spring-mass model for running and hopping," *Journal of Biomechanics*, vol. 22, pp. 1217–1227, 1989.

[3] T. A. McMahon and G. C. Cheng, "The mechanics of running: How does stiffness couple with speed?" *Journal of Biomechanics*, vol. 23, pp. 65–78, 1990.

[4] C. T. Farley and D. P. Ferris, "Biomechanics of walking and running: From center of mass movement to muscle action," *Exercise and Sport Sciences Reviews*, vol. 26, pp. 253–285, 1998.

[5] T. F. Novacheck, "The biomechanics of running," *Gait and Posture*, vol. 7, pp. 77–95, 1998.

[6] A. M. Grabowski and H. M. Herr, "Leg exoskeleton reduces the metabolic cost of human hopping," *Journal of Applied Physiology*, vol. 107, pp. 670–678, 2009.

[7] C. T. Farley and O. Gonzalez, "Leg stiffness and stride frequency in human running," *Journal of Biomechanics*, vol. 29, no. 2, pp. 181–186, 1996.

[8] A. A. Biewener, "Muscle function in vivo: A comparison of muscles used for elastic energy savings versus muscles used to generate mechanical power," *American Zoology*, vol. 38, pp. 703–717, 1998.

[9] R. M. Alexander, "Energy-saving mechanisms in walking and running," *Journal of Experimental Biology*, vol. 160, pp. 55–69, 1991.

[10] G. A. Cavagna, N. C. Heglund, and C. R. Taylor, "Mechanical work in terrestrial locomotion: Two basic mechanisms for minimizing energy expenditure," *American Journal of Physiology Regulatory, Integrative, and Comparative Physiology*, vol. 233, pp. 243–261, 1977.

[11] T. A. McMahon, *Muscles, Reflexes, and Locomotion.* Princeton, NJ: Princeton University Press, 1984.

[12] D. P. Ferris, "Running in the real world: Adjusting leg stiffness for different surfaces," *Proceedings Biological Sciences*, vol. 265, pp. 989–994, 1998.

[13] A. E. Kerdok, A. A. Biewener, T. A. McMahon, P. G. Weyand, and H. M. Herr, "Energetics and mechanics of human running on surfaces of different stiffnesses," *Journal of Applied Physiology*, vol. 92, pp. 469–478, 2002.

[14] D. P. Ferris, K. Liang, and C. T. Farley, "Runners adjust leg stiffness for their first step on a new running surface," *Journal of Biomechanics*, vol. 32, pp. 787–794, 1999.

[15] B. Wiggin, S. H. Collins, and G. S. Sawicki, "An exoskeleton using controlled energy storage and release to aid ankle propulsion," in *Proceedings of the 12th Biannual International IEEE Conference on Rehabilitation Robotics*, 2011.

[16] M. S. Cherry, S. Kota, and D. P. Ferris, "An elastic exoskeleton for assiting human running," in *Proceedings of the ASME 2009 International Design Engineering Technical Conferences & Computers and Information in Engineering Conference*, 2009.

[17] G. A. Cavagna, "Force plates as ergometers," *Journal of Applied Physiology*, vol. 39, no. 1, pp. 174–179, 1975.

[18] J. M. Brockway, "Derivation of formulae used to calculate energy expenditure in man," *Human Nutrition Clinical Nutrition*, vol. 41, pp. 463–471, 1987.

[19] J. He, R. Kram, and T. A. McMahon, "Mechanics of running under simulated low gravity," *Journal of Applied Physiology*, vol. 71, pp. 863–870, 1991.