Investigation of dielectric elastomer human energy harvesting to reduce knee joint torque deviation due to bracing

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Abstract
Introduction: The use of a smart electromechanical material, dielectric elastomer, is investigated for the development of an active bracing technique, which modifies the stiffness and damping of the knee brace during energy harvesting so as to reduce knee joint torque deviation during late swing in braced walking.

Methods: The bracing technique considered involves a dielectric elastomer energy harvesting cycle, which activates only when the knee flexor muscles are contracting eccentrically during late swing. The brace reduces the leg extension deviation during late swing in braced walking by transforming a portion of the mechanical stored energy into electrical energy, reducing the required internal work performed within the body.

Results: Simulated behavior of the dielectric elastomer brace worn across the knee joint demonstrates that when properly activated, the dielectric elastomer brace's reduction in stiffness and increase in damping minimize the added energy expenditure of knee joint bracing during late swing.

Conclusions: The modeling results demonstrate the effective application of a soft, circumferential, dielectric elastomer energy harvesting knee brace, which utilizes the changes in the dynamic behavior of the knee joint occurring during energy harvesting in order to reduce the added demand placed on the knee joint under braced conditions.

Keywords
Energy harvesting, active bracing, knee joint bracing, smart materials

Introduction
Research has long demonstrated that bracing alters the joint torque and power patterns on the knee during walking and running¹ producing significant deviations of the muscle activity,² resulting in increases in the metabolic exertion of the wearer.³ Even healthy, asymptomatic wearers experience alteration of their exertion due to bracing. In the case of steady-state treadmill running, for example, bracing results in increased metabolic rates of 3–6% above those of non-braced running.² Also, muscle activation for knee and ankle flexor/extensor muscle groups is significantly altered due to the presence of bracing.³ This is one of the main reasons pro athletes have varying opinions on the use of prophylactic bracing to prevent injuries during active sports. With the development of energy harvesting components using smart materials, there may be potential to negate the effect of the brace on gait and knee joint torque demand for wearers who use soft, prophylactic sleeve braces while engaging in sustained activity, potentially reducing the metabolic demand of bracing.

Biomechanical energy harvesters have the ability to generate and store energy created by human motion during the gait cycle. Early researchers found the eccentric work at the knee joint to have the most promising potential.⁴ With the recent developments of low powered, small-scale electronics, it is possible to provide a...
means to carefully control the behavior of increasingly sophisticated controls on smart, wearable devices. These smart wearable devices are able to interact with the wearer during use. The ability to perform carefully controlled energy harvesting during movement has allowed for the development of beneficial energy harvesting devices, which transform excess mechanical motion into electrical energy. This electrical energy can be stored for later use in a variety of wearable systems. Promising examples of beneficial energy harvesting include the PowerWalk, a commercially available knee joint harvester,5 and a transfemoral prosthesis, developed by Andrysek et al.6 Both of these devices were developed using a series of gears and clutches to engage and disengage a small generator as desired. The end result is an effective, yet bulky and noisy electromechanical energy harvester, which can be designed to partially offset the metabolic cost of wearing the mechanism. Applying these concepts to a state of the art, wearable electromechanical elastomer material, produced a design that harvests electricity while strategically modifying the behavior of the soft, circumferential, sleeve knee brace. This provides the ability to actively control the stiffness of the material utilized in soft prophylactic knee braces, improving the performance of athletes using lower extremity bracing to supplement healthy muscle function.

The purpose of the research presented here is to propose, through bench-top testing and musculoskeletal modeling, the potential layout and timing for active knee bracing using an electro-mechanical smart material. The smart material selected is the dielectric elastomer (DE), a class of soft, polymeric composite materials, which can be electrically activated to act as an actuator, sensor, or energy harvester. Using DE, energy transfer can be controlled as a function of geometry, electrical loading conditions, and material properties, affording the opportunity to develop a knee brace that can be tailored to the wearer to provide the appropriate stability throughout the gait cycle. This paper describes the recommended orientation and timing for a soft, circumferential, sleeve-type knee brace, which harvests energy in a manner that reduces the demand of the knee joint torques, compared to a passive circumferential sleeve knee joint with a similar constant stiffness during stance.

Methods

The study presented here begins with an overview of the proposed DE harvesting circumferential sleeve brace design and energy harvesting scheme, including a description of the knee joint analysis utilized to establish the orientation and timing of the bracing device. This is followed by a description of the simulation and modeling performed to evaluate the effect of energy harvesting on the gait cycle response of the wearer through estimates of knee joint torque and power demands as a result of DE energy harvesting within the sleeve.

Application of active DE energy harvesting to a soft circumferential, sleeve brace

The proposed design is a soft circumferential sleeve brace that contains electrically active material in only specific portions of the sleeve. The orientation and timing of the DE harvesting circumferential sleeve brace is dictated by knee joint kinematics and kinetics. First, the relative motion of the DE energy harvester directly affects the energy harvesting capability of the device. Therefore, the kinematics of the gait motion are taken into consideration in the determination of the location and timing of the activation of the device. Second, the musculoskeletal system has a dynamic response at the knee joint to DE energy harvesting, which is directly related to the kinetics of the gait motion. Therefore, incorporation of both of these two interrelated aspects is critical for the development of a timing profile for the smart DE brace.

Proposed design of knee joint DE energy harvesting cycle. Dielectric elastomers are comprised of soft, highly elastic dielectric material, with compliant conductive electrodes coating both sides. This compliant fabric behaves as a variable capacitor that is highly influenced by the electrostatic Maxwell stresses formed when a voltage is placed across the elastomer. The DE material used in this study is a material produced by Danfoss PolyPower A/S, which is comprised of a unidirectional corrugated silicone dielectric film with vacuum sputter deposited metallic electrodes.

The energy harvesting capacity of a DE energy harvester is directly related to the stretch that it experiences, therefore, the knee joint angle profile plays an important role in determining how to coordinate the timing of electrical loading of the device. During mid to late swing phase, the knee joint experiences a very large transition from flexion to extension. When the DE material contained within the sleeve brace is placed across the front of the knee joint as shown in Figure 1(a), the device can be controlled to harvest energy when the greatest relaxation of the DE material occurs during the late swing phase, as demonstrated in Figure 1(b). In this configuration, the dielectric elastomer strip is oriented within the sleeve so that it is located across the patella. The unidirectional DE undergoes uniaxial stretch along the leg, while undergoing a constant charge energy harvesting cycle.

This proposed location for the DE results in an energy harvesting cycle that begins with the knee
extended during the early stance phase. In this position, the DE device is at its minimum stretch. When the knee joint flexes during late stance, a uniaxial mechanical stretch is applied to the DE, and the capacitance increases based on the elongation and thinning of the dielectric film. When the knee joint is at maximum flexion, a constant charge is placed across the electrodes of the device, and the induced Maxwell stress causes the dielectric elastomer to thin even more, and the surface of the electrodes to expand, resulting in a slight increase in the capacitance of the DE. As the lower limb continues through late swing phase, the knee joint extends, and the DE returns to nearly its original shape, causing the capacitance of the DE to decrease, and the voltage across the DE to increase. At full extension, just before heel strike, the capacitor is discharged, allowing the original voltage, plus the additional charge generated due to the Maxwell stress, to flow into an external storage device or return to the circuit to continue powering the system.

The application of the Maxwell stress at full flexion causes a reduction in the stress/strain behavior of the braced knee during early swing phase, reducing the apparent stiffness of the sleeve. Additionally, the unrecovered mechanical stretch energy, which is converted into electrical energy at the end of the gait cycle results in a reduction in the total energy which must be removed from the lower leg. The behavior of the active DE material is transmitted from the front of the knee to the leg through the soft circumferential sleeve, specifically when the knee flexor muscles are contracting eccentrically to slow down the lower leg during late swing. In this way, it is proposed that the knee joint with an active DE brace will have torque and power demands closer to the unbraced knee than possible with a passive circumferential sleeve brace.

**Figure 1.** (a) Circumferential sleeve brace with location of dielectric elastomer active material shown in black. (b) Area of greatest DE stretch shown in red as the knee joint moves from flexion to extension during late swing.

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**Knee joint kinematics and kinetics during gait cycle and beneficial energy harvesting.** The response of the wearer to the DE harvester is directly related to the kinetics of walking, and the determination of the appropriate location and timing of the DE generator is critical to minimizing gait deviations resulting from the harvesting process. For this reason, development of the appropriate timing for energy harvesting utilizes the knee joint intersegmental power curves seen in Figure 2. Observation of the power curve indicates that the majority of the joint power occurs from late stance through the swing phase. During the pre-swing phase in late stance, over half of the knee flexion required for foot clearance occurs prior to the toe lifting off. Knee flexion in this phase of gait is passive in nature as the contralateral limb accepts the weight of the body. During early swing it is the action of the hip flexors which provide the power for knee flexion and swing limb advancement. As the shank and foot are decelerated during late swing, the knee flexor muscles contract eccentrically, requiring internal work to be performed within the body. During this deceleration, the muscles must expend energy to absorb and dissipate kinetic energy in order to slow down the lower limbs and control the extension velocity of the lower leg. The DE harvester is worn as a soft, sleeve brace across the knee joint and when activated, it will convert small amounts of mechanical energy into stored electrical energy, thereby assisting in the dissipation of the kinetic energy of the lower leg during swing. When the kinetic behavior of the knee is synchronized with the DE harvester, it is possible to remove energy from the knee and induce beneficial damping at the specific point in the gait cycle when damping is needed, thereby reducing the metabolic cost of wearing the brace. The focus of this study did not address the impact of the DE harvester at the knee during initial swing due to the complex and coupled nature of hip and knee flexion during early swing.

By incorporating the behavior of DE energy harvesters with lower limb kinematics and kinetics during...
walking, electrical loading patterns for a DE generator placed across the front of the knee joint are developed which selectivity energize the device such that beneficial energy harvesting occurs. This coordination is demonstrated in Figure 2 where the DE activation during late swing is highlighted. Using this orientation and electrical loading pattern, the DE device can effectively reduce the deviation of the knee joint torque as compared to the nonbraced gait during late swing phase.

Modeling of gait response to active DE energy harvesting

A three-part computer model is used to simulate the behavior of an active DE knee brace worn during walking. First, the change in the damping and stiffness of the knee joint during swing phase due to the presence of an activated DE brace is simulated using a test stand and joint oscillation model. Secondly, musculo-skeletal gait analysis is used to model the behavior of the knee while walking with and without a brace. Finally, a joint kinetics model is used to combine the estimated damping and stiffness coefficients of an active DE brace to the results of the musculo-skeletal model. The results of these models are used to estimate the effect of the active DE energy harvester on knee joint torque and power demand.

Simulation of the behavior of the activated DE harvester during swing phase. The nonlinear dynamic response of the knee joint to DE energy harvesting is estimated using a comparison of the simulated rotation of the knee joint during swing phase with the oscillatory motion of a simple pendulum. Using the assumption that the knee joint during swing phase behaves similarly to a simple pendulum, its motion is modeled by a one degree of freedom, angular equation of motion, 

$$\dot{\phi} = -k(\phi - \phi_0) - c\dot{\phi} + M_M$$

where $k$ and $c$ relate to the stiffness and damping of the knee joint and $M_M$ is the moment due to the Maxwell stress, which is only nonzero during charging. For this study, $k$ and $c$ were selected based on empirical measurements of the knee joint test stand, and the Maxwell stress is calculated based on the configuration of the DE: $M_M = \varepsilon D \frac{V^2}{2}$. The stretch ratio, $\lambda$, is a linear function of the angle based on the end-point stretch ratios. The terms $w$ and $t$ are the initial width and thickness of the DE material, $D$ is the perpendicular distance from the measurement location to the knee joint, $\varepsilon$ is the permittivity, and $V$ is the charge voltage. Numerical simulation of the angular displacement is performed in MATLAB using ODE45, and the resulting damped oscillation waveforms are used to estimate the stiffness and damping coefficients using logarithmic decrement of simple harmonic motion. In this manner, the stiffness and damping coefficients are found for increasing charging voltages, corresponding to rising levels of energy harvesting. These stiffness and damping values are stored in tabular format for use with the musculo-skeletal modeling of the knee joint behavior during active bracing described in the section “Simulation of the effects of an active DE brace on walking”.

Simulation of the effects of an unactivated DE brace on walking. The second component of the modeling utilizes the musculo-skeletal platform OpenSim 3.0 (developed by the National Center for Simulation in Rehabilitation Research) to develop curves for the knee joint torque and power demand for an unbraced knee and a knee with an unactivated DE circumferential sleeve brace. These two curves are used to establish the bounds within which the active DE brace should be designed to charge.

The OpenSim simulations are performed using a scaled measurement based model of the lower body (Gait2354_Simbody) and a set of marker data for a single, representative test subject (data available at OpenSim website). The basic biomechanical model comprises bodies, joints, forces, and markers. The muscles are individually modeled as specially defined forces.

Figure 2. Knee joint intersegmental power during walking gait cycle: yellow indicates eccentric muscular action of the knee joint muscles, and the blue shaded portion relates to late swing when the DE energy harvester would be activated.
which have prescribed attachment points and dynamic behavior. Figure 3(a) and (b) shows the muscles associated with extension and flexion of the knee. Flexor muscles incorporated into the model include the biceps femoris, semitendinosus, semimembranosus, sartorius, gracilis, lateral, and medial gastrocnemius. Extensor muscles incorporated into the model include the rectus femoris, vastus lateralis, vastus intermedius, and vastus medialis.

The scaled model utilizes the distances from the statistically measured marker locations (which are associated with externally visible anatomical landmarks) to the internal anatomical coordinate systems related to each segment. See Figure 3(c) for the coordinate systems related to the left hip, knee, and subtalar joints, where the $z$-axis (green) is perpendicular to the sagittal plane and the $y$-axis (yellow) is oriented along the represented bone segment.

The movement marker data are transformed through inverse kinematics to determine the relative movement between segments, and hence the relative joint angles between them. The flexion/extension of the knee joint within the sagittal plane (rotation in the $z$-axis) is of specific interest in this investigation. Published marker location and measured ground reaction force data are used to perform inverse dynamics calculations resulting in joint reaction forces and torques related to each limb segment. This process generates the characteristic kinetic profile of each lower limb joint.

These profiles provide the baseline for the analysis of the effects of the energy harvesting DE located at the knee joint. In order to compare the energy expended by the knee with active DE harvesting to that of a normal knee, the behavior of the DE is coupled to the measured moment in OpenSim using a modeling component called a bushing force. This component is specifically designed to model forces, which are a function of both displacement and velocity, similar to the perceived effect of the mechanical behavior of the braced knee: \[ \tau_{DE} = k_r \theta + c_r \dot{\theta}, \]

where $k_r$ and $c_r$ are the rotational stiffness and damping coefficients, respectively.

The bushing force component consists of a force that is applied at a specified joint and is defined in terms of linear and rotational stiffness and damping, which must be specified in all three directions. In modeling the DE harvester, the bushing force component is positioned at the knee joint and it is defined based on the relative motion between the femur and the tibia, with full extension occurring at $\theta = 0$, hyperextension at $\theta < 0$ and flexion at $\theta > 0$. In defining the active DE harvester, the translational forces are all defined as zero. Rotational stiffness and damping terms are applied to the bushing force model in the $z$ direction, corresponding to the flexion and extension motion of the knee. The values used for the rotational stiffness and damping terms defined in the bushing force are $\Delta k_r$ and $\Delta c_r$ for the unactivated brace with zero energy harvesting found using the damped swing phase simulation (Table 1). This additional force related to the effect of an unactivated DE brace provides a comparison of the joint torque for the unbraced knee and the braced knee, and used to define the operating parameters for an active DE knee joint brace that is able to harvest energy, while resulting in gait demand that is closer to the unbraced demand.

**Simulation of the effects of an active DE brace on walking.** The final stage of the modeling uses the torque and power demand curves to estimate the
Table 1. Modeled change in damping coefficient ($\Delta c_r$) and rotational stiffness ($\Delta k_r$) due to increase in energy harvesting resulting from charging conditions for the active DE brace.

| Energy harvested (J) | $\Delta c_r$ (N-m s/rad) | $\Delta k_r$ (N-m/rad) |
|----------------------|--------------------------|------------------------|
| Unactivated DE brace | 0.0205                   | 1.189                  |
| 0.05                 | 0.0302                   | 1.176                  |
| 0.11                 | 0.0428                   | 1.165                  |
| 0.30                 | 0.0909                   | 1.110                  |
| 0.60                 | 0.1806                   | 1.039                  |
| 0.99                 | 0.2695                   | 0.687                  |
| 1.22                 | 0.2348                   | 0.405                  |

The effect of active DE energy harvesting on the wearer. The effect of an active DE energy harvesting brace is modeled in MATLAB by applying the inverse kinematics and dynamics results with the discrete mechanical loading, due to the Maxwell stress generated during charging and discharging. This model simulates the knee joint intersegmental torque due to active DE bracing, where the DE is charged only during the late swing phase as a summation of the torques caused by the stiffness and the damping induced by the device when active with the unactivated DE brace:

$$t_{DE\ brace} = t_k + t_h + t_{unactive\ brace}.$$  

Based on the location of the DE material, across the front of the knee joint, the DE harvester behavior is modeled in two parts: the uncharged behavior of the DE harvester, related to the mechanical properties of the device, and the charged behavior of the device, related to the electromechanical coupling during energy harvesting. The transition between the unloaded and loaded conditions is determined based on the onset of the extension portion of the swing phase. When the knee joint flexion is at its peak, the stiffness and damping coefficients for the model reflect the behavior of the charged device due to electrical loading. At the end of the swing phase, before heel strike occurs, the stiffness and damping coefficients return to their original values, corresponding to the uncharged DE device.

This joint dynamics model provides a means to quantify the energy requirement at the knee joint to operate a DE harvester. The intersegmental power profile is found using the velocity and torque profiles. The velocity profile is calculated by differentiating the original joint angles, while the torques are known from the prior inverse dynamics calculations. Multiplying each point on the velocity and moment curves, together over the gait cycle, results in a power profile. By integrating the power profile during the swing phase, the energy absorbed by the knee during extension is determined. This energy term is used to characterize the effect of different levels of DE energy harvesting; the more energy harvested from the system, the less energy expended by the muscles at the knee to slow down the motion of the leg as it swings through its pendular trajectory. Extrapolation of the behavior for DE materials with increased electrical performance is used to assess the efficacy of the active bracing scheme presented based on the feasibility and benefit of selective DE energy harvesting during the swing phase.

**Results and discussion: Modeled effects of energy harvesting on knee joint torques**

In the following section, the results of each of the three computer simulations are discussed, focusing on the modeled effects of an active DE soft, circumferential, sleeve brace on the torque and power demand at the knee joint while walking.

**Results from simulation of damped swing phase knee joint**

The joint oscillation model serves to estimate the nonlinear effect of energy harvesting on the mechanical properties of the knee joint. The modeled effects of energy harvesting can be seen in Table 1, which displays the change in knee joint rotational damping and stiffness due to energy harvesting based on the second-order dynamic behavior of the lower limb during late swing. These results demonstrate that as energy harvesting increases, the perceived damping of the knee joint (due to the sleeve) increases, while the perceived rotational stiffness decreases.

The results demonstrate that the amount of energy harvested from a DE device has a direct effect on the behavior of the active DE brace, underpinning the assertion of this research that it is possible to develop an active DE energy harvesting sleeve brace. The material properties of the brace can be changed to alter the perceived knee joint mechanical properties during the gait cycle. This further demonstrates the potential for developing an active DE brace that provides a higher level of stiffness during the stance phase, while allowing for reduced stiffness and increased beneficial damping during the swing phase.

**Results of simulating the effects of an unactivated DE brace on walking**

A comparison of the intersegmental torque demand at the knee due to normal walking with and without an unactivated circumferential DE knee brace is developed using the OpenSim musculo-skeletal model and is presented in Figure 4.

The effect of the unactivated DE brace is most pronounced during the eccentric demand of the knee joint...
throughout the swing phase. The increase in the eccentric work required by the knee flexors in a braced knee is caused by the additional stiffness of the unactivated DE brace acting on the knee joint. The increased stiffness creates an additional knee extension torque, requiring greater work to be performed by the eccentrically contracting knee flexor muscles in order to slow down the knee joint in the preparation for heel strike. These modeling results, in conjunction with the kinematic and kinetic analysis of the gait cycle, indicate there is a need for an active DE brace that could activate during the mid to late gait cycle as highlighted in Figure 4. The goal of this proposed active damping scheme would be to harvest small amounts of energy while reducing the additional late swing torque requirement associated with wearing a passive circumferential knee brace.

In addition to confirming the proposed energy harvesting scheme, these joint torque profiles developed using inverse kinematics and dynamics, define the baseline joint torque values. They identify that the modeled response to a circumferential sleeve brace is directly related to the stiffness and damping parameters through superposition of the torque induced by the change in damping and stiffness with the unbraced knee joint torque.

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Results of simulating the effects of an active DE brace on walking

The effect of an active DE energy harvesting brace on the knee is assessed through comparison of the torque and power demand of the knee joint undergoing active DE bracing with that of an unbraced knee. The modeled behavior of the knee joint during energy harvesting is observed in Figure 5, which provides a close-up of the torque and power demand during the late swing phase activation of the DE brace. This graph demonstrates the braced knee joint response to increasing energy harvesting imposed by changes in the active bracing electrical parameters compared to the original unbraced knee. The black solid and dash/dot lines correspond to the unbraced knee and the unactivated DE braced knee, respectively. The other lines demonstrate the modifications to the knee joint demands as energy harvesting increases. For the energy harvesting patterns shown, the knee joint torque and power demands are seen to be closer to the original unbraced knee joint demands at the higher energy harvesting configuration.

As expected, the change in the mechanical behavior of the knee as a result of energy harvesting (Table 1) directly influences the joint dynamics. The addition of the brace across the knee joint ($E_{k} = 0$) causes the negative torque to increase throughout the eccentric loading of the knee joint during late swing. The larger negative energy at the joint indicates that more work must be done by the muscles to absorb energy while controlling the motion of the lower limb while wearing a traditional brace. This detrimental effect is due to the increased stiffness imposed by the brace on the knee joint.

Based on these results, there is a direct relationship between the amount of energy harvested and the response of the knee during the swing phase extension. As the stiffness of the brace decreases and the internal damping of the DE brace increases due to DE energy harvesting, the eccentrically loaded muscles near the knee are required to perform less negative work in order to control the motion of the lower leg in preparation for heel strike. This results in torque and power demands, which more closely match the profile of the unbraced knee, potentially decreasing the gait deviation resulting from the original bracing. The negative additional energy expenditure for the highest energy harvesting case demonstrates that it is possible for enough energy to be removed from the knee as a result of DE harvesting during the swing phase extension to reduce the energy expenditure of the muscles below the
original energy expenditure for the knee alone. While this does not guarantee that the overall biomechanical energy expenditure for the entire gait cycle will be less, it does indicate that there is potential for harvesting conditions in which the energy harvested and overall energy expenditure can be optimized either to reduce gait deviation or to reduce the metabolic expenditure of the wearer.

**Conclusion**

The purpose of this research is to develop a model of the damping effects of the DE energy harvester on knee joint kinetics and apply it to the analysis of an active DE soft, circumferential, sleeve knee brace. The objective is to demonstrate that proper timing and control of DE harvesting can be used to actively modify the stiffness and damping of the sleeve, in order to reduce the gait deviation caused by bracing during the late swing phase. To this end, a musculoskeletal model of the lower body is utilized to compare the knee joint torque generated during the gait of a healthy subject without a brace to one wearing an electromechanical knee brace. Using this model, a DE energy harvesting scheme is proposed such that the changes in damping and stiffness due to energy harvesting reduce the joint torque demand at the knee due to the presence of a knee brace during late swing.

In order to estimate the effects of the DE energy harvesting process on the biomechanics of the knee, the knee joint behavior without a DE energy harvester is compared with the expected response of the knee joint during DE energy harvesting, using a three-part model involving joint oscillation, musculo-skeletal modeling and joint kinetics. The joint oscillation model utilizes a simulation of the shank and foot as an oscillating pendulum to estimate the stiffness and damping coefficients of the knee joint during DE energy harvesting. The musculo-skeletal model uses an open source, biologically-based platform to simulate the knee joint behavior of an unbraced versus a passively braced knee. Combining the results of these two models, a final joint kinetics model estimating the changes in knee joint torque and power demands due to the DE energy harvesting is presented.

Analysis of the effect of increasing energy harvesting demonstrates that beneficial energy harvesting can be achieved such that the overall increase in the energy requirement is reduced as the energy harvested increases. This is done through strategic timing of the DE energy harvesting to produce improvement to the stiffness and damping of the brace during late swing. The results of this analysis are significant, as they confirm that there is potential to use soft, wearable active electromechanical devices for beneficial energy harvesting, and that it is possible to design a DE energy harvesting knee brace.

**Figure 5.** Torque and power curves of DE harvester for increasing harvested energy ($E_h$).
that reduces the deviation of the knee joint kinetics often experienced under traditional braced conditions.

Evaluating the bio-kinetic effects of active bracing based on a mechanically simulated model does have limitations. First, this model is limited to the behavior of one specific active bracing configuration, and it assumes a linear response of the knee joint to the changes in brace properties. There is the potential for many other configurations, which may result in a non-linear response of the knee joint to changes in material properties. Additionally, in this evaluation, it is assumed that joint torque demand at the knee alone is an accurate predictor of gait deviation. Modeling of joint kinetics at other joints, or modeling other predictors, may be necessary to improve the fidelity of the overall gait deviation assessment. Finally, it has been assumed that the timing of the DE energy harvester can be perfectly controlled. It is understood that the control of the timing is not trivial and would require the identification of gait cycle time as a function of joint angle through a technique such as measurement of the DE capacitance. Therefore, further modeling and laboratory testing related to each of these issues is necessary to assure the safety and efficacy of DE energy harvesting circumferential sleeve bracing.

It is also important to note that the modeling presented here relates to a low power application of DE energy harvesting in a soft, circumferential sleeve brace. This type of brace has the intended purpose of addressing knee instability due to mild pain or weakness in a typically healthy user, not to address significant muscular deficiencies. For a device such as this, the small changes in the material’s behavior during charging has a positive effect on the wearer’s use of this device, as the variable stiffness allows for the brace to be designed with more support during stance and less stiffness during later swing. This allows for less eccentric contraction demand of the knee flexor muscles during late swing motion. Additionally, the active DE material embedded in the soft circumferential sleeve brace presents opportunities for auxiliary applications, such as measurement and storage of joint angles or gait deviation. Also, once the challenges described related to the low power operation of a soft, circumferential sleeve-type brace have been addressed, additional investigation into the application of higher power DE energy harvesting schemes to devices, which may require additional power such as stiff, hinged orthotics, or powered prosthetics is also recommended.

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HL researched literature and conceived the study. TR provided technical details related to clinical biomechanics. HL wrote the first draft of the manuscript. All authors reviewed and edited the manuscript and approved the final version of the manuscript.

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