SLIDE FORUM—PERFUSION AND CIRCULATORY ASSISTANCE TECHNIQUES 5

A Gait-Powered Autologous Battery Charging System for Artificial Organs

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The quality of life of patients relying on electrically powered artificial organs is currently restricted by the limited energy availability provided by portable batteries. As these patients become increasingly ambulatory, and are developing more active lifestyles, this limitation grows more apparent. Coincidentally, these patients may themselves be capable of generating electrical power as a consequence of their physical activity. Extraction of this latent autologous energy could, in turn, be used to augment charging of internal batteries—thus untethering the patient from external power for extended periods of time. In this study, the viability of deriving energy associated with natural human ambulation has been evaluated. The kinematic components of gait were evaluated to identify the largest useful forces and moments that may be harnessed as an energy source, while presenting minimal "perceived" work for the patient. It was found that the ground reaction forces associated with the heel strike and toe-off phases of the gait represent the greatest potential for usable energy. This study uses a piezoelectric array within the midsole of the shoe for the conversion of mechanical to electric energy. This power could then be easily coupled in tandem with existing transcutaneous transformers for augmenting or temporarily replacing external power sources. *ASAIO Journal* 1995; 41:M588–M595.

An ongoing consideration in the development of permanent artificial organs is the choice of energy supply. Although it may be tolerable to transmit power percutaneously during the investigational stage of these devices via pneumatic tubes or electrical cables, this limitation must be overcome for systems to be totally permanent and portable. Implanted artificial organs that require significant power (from 5 to 20 W), such as circulatory support pumps, will therefore demand energy storage on the order of 40–160 W-hr to achieve 8 hr of autonomous operation without intervention. Such a battery pack, built using existing nickel–cadmium techology, would result in a 12 volt battery system weighing over 1.3 kg (57.6 W-hr) using D-sized cells, or 3.9 kg (160.8 W-hr) using F-sized cells. Even with the latest battery technology that could potentially reduce this battery mass, there remains the desire to miniaturize the batteries further, and increase their operating range. This need has become increasingly apparent as artificial heart patients become more active. Thus, the perceived limitation imposed by a limited power supply becomes more acute as patients are rehabilitated.

The current study was conducted to evaluate the feasibility of extracting useful energy from ambulation to provide supplemental power for operating electrically powered artificial organs, with particular emphasis on ventricular assist devices. The study investigated the most likely modes of extracting energy, and considers several design factors for converting mechanical energy from ambulation.

Materials and Methods

Power Requirement

The hydrodynamic power supplied by the heart to circulation ranges from approximately 1.2 W at basal cardiac output for a normotensive subject, to approximately 5 W at maximum exercise. The efficiency of electric pumps corresponding to physiologic pressures and flows is in the range of 10–50%. Depending on the flow rate and pump design, these devices can consume between 3 to 40 W of electric power. Additional inefficiencies associated with power conditioning and transcutaneous energy transmission may increase this range to 4–65 W. The most efficient systems to date, such as the Nimbus AxiPump, the NASA/Baylor axial flow ventricular assist device (VAD),[1] the Jarvik 2000,[2] and the VAD described by Okamoto *et al.*,[3] draw between 7 and 13 W. Adding a transcutaneous energy transmission system would increase this capacity requirement to 9.5–16.5 W. Based on the speculation that future generation systems will further improve their overall efficiency, the targeted power output for the autologous generator was taken to be 10 W at a moderately paced gait.

Analysis of Gait: Recoverable Energy

For a given cadence, the power availability is estimated by the dynamics of the lower extremities. Ranges of motion, forces, and moments associated with a normal gait pattern

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were evaluated for their potential as power sources. Treating the body as linked segments, each of the leg segments and joints were assessed during the stance and swing phases of gait. Large moments, forces, or angular displacements that could occur were of particular interest because they would represent the most likely candidates. This analysis was based on kinematic and dynamic data published by Winter[4] for a 56.7 kg person walking at a cadence of approximately 1 step/sec. These data were used to derive available mechanical power associated with each leg segment and joint.

Because mechanisms used to harness mechanical energy occurring at the joints would most likely be inserted across the joint, relative angles and angular velocities between the leg segments during gait were used to calculate joint external power. During the stance phase, the femur passes through a range from 10° extension to 30° flexion at the hip joint, whereas in the swing phase it ranges from 0° to 30° flexion. Similarly, the knee passes through 0° to 40° in the stance phase, and 0–60° in the swing phase. Smaller ranges of motion are found at the ankle, with a range of 15° dorsiflexion to 20° plantarflexion during stance, and 0–20° plantarflexion in the swing phase.

Moments occurring between the thigh and shank at the knee reach a peak of approximately 35 N-m during, and immediately after, heel contact. Rotational net muscular power at the knee instantaneously rises to slightly more than 10 W during the swing phase and immediately after the heel contact in the stance phase. A similar analysis of the ankle yields a peak 90 N-m moment during the stance phase. Net muscle moments at the ankle produce a power profile that results in a single peak per cycle of approximately 37 W. Similar calculations for the hip, possessing a peak moment of 54 N-m, yield an approximate mean power of 77 W. Although the power associated with the knee and ankle movements during gait are of the order of magnitude required, harnessing this power would place the patient at a deficit, requiring extensive exertion to maintain ambulation.

Preconditioning the patient before using the device would be required to offset deficits in joint power or, alternatively, a slower ambulation speed must be accepted by the patient. Furthermore, devices necessary to harness these joint power sources are expected to be cumbersome, adding excessive weight and external resistance to the patient during ambulation. These devices, which may be visually perceived as "special equipment" to aid the patient, will negatively affect patient acceptance.

In addition to the relatively large increase that would be required for adequate joint power development, the rotational ankle power peak is followed by periods of little or no net muscle power, because of an extended period of minimal or no ankle moment. The power associated with the ankle and knee was therefore dismissed from further consideration as a viable power source for this application.

In contrast to the hip, knee, and ankle, which are typified by mechanically efficient movements, the energetics of the foot are commonly associated with a significant proportion of dissipated energy. The ambulatory patient is already accustomed to energy extraction by the shoe (as contrasted to the impediment of wearing a brace crossing a joint, for example). During foot impact, a portion of the work performed by the plantar surface of the sole is transformed into

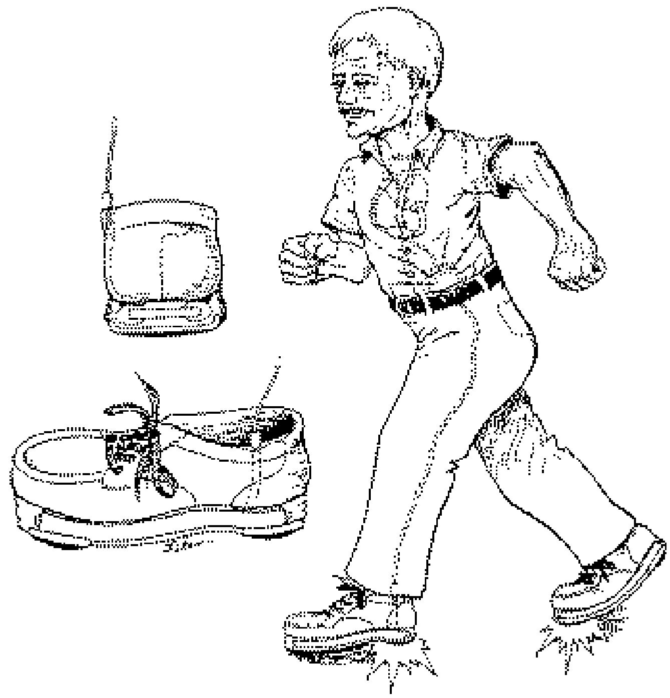

Figure 1. Conceptual shoe generator system. Electric energy generated in midsole is transmitted to the external battery charging unit through a remote port connection.

strain energy that is temporarily stored in the elastically deformed sole, while the remainder is dissipated as heat due to viscous losses.[5] With the release of foot loading from the sole, the strain energy is returned by the sole to the foot in elastic recovery, with a portion of the energy again lost through heat dissipation. A generator situated in the shoe should therefore aim to recover most of the energy that would normally be dissipated, as well as a portion of the recovered strain energy. With higher levels of power available at the foot, siphoning off the required power will result in a smaller percent increase in the work input of the patient, compared to using the leg joints. This would thus impose the minimum loss of efficiency toward propulsion, as well as limit the patient's perception of additional work.

Shoe Generator Design

A shoe generator was designed in consideration of several mechanical, electric, and ergonomic factors. The principal components of the system, shown schematically in **Figure 1**, consist of the piezo transducer, mechanical coupling, and power conditioning.

The potential power output of such a shoe generator is principally restricted by the ergonomic factors governing deliverable power to the midsole. To evaluate further the limit of deliverable power, ground reaction forces occurring at the foot were analyzed for their use in producing translational mechanical power. As shown in **Figure 2a**, the force profile for a moderately paced walking step is characterized by a double peak at approximately 1.2 times the body weight, occurring as a result of the heel strike and forefoot push-off.[4,6]

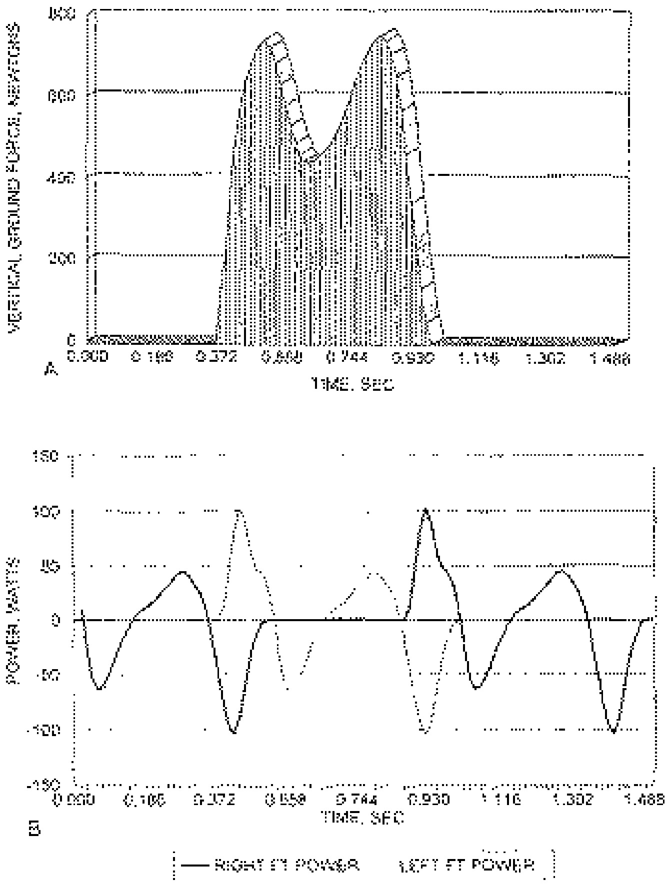

Figure 2. (A) Typical ground reaction force for 70 kg subject (adapted from Winter D: *Biomechanics and Motor Control of Human Movement*, 2nd ed. New York, John Wiley & Sons, 1990.). **(B)** Estimated power generated at shoe sole, based on 1 cm total displacement.

These forces can be used in the production of external work by allowing the sole of the shoe to deform. Accordingly, the average power transferred to the midsole is given by:

$$\dot{W}_s = \frac{1}{2}(F_h\delta_h + F_t\delta_t)\cdot f \qquad (1)$$

where F_h and F_t are the heel and toe forces, respectively, δ_h and δ_t are the associated displacements, and f is the cadence (i.e., frequency of ground strike for each foot). Based on typical measurements observed in running shoe midsoles, it was assumed that 10 mm would represent a maximal allowable displacement at this pace. The result of this analysis indicates alternating 6 N-m peaks of work being performed by each foot during the stance phase. The dynamic power associated with ground reaction forces is shown in **Figure 2b**, which indicates that instantaneous power levels as high as 100 W are attainable during heel contact. This corresponds to a total mean power of 8.2 W per foot under moderate gait (1 step/sec), for a typical 70 kg person ($F_t = F_h = 823$ N), or a total of 16.4 W. A similar calculation for running ($F_t = F_h = 1500$-2000 N, $\delta = 20$ mm, $f = 3s^{-1}$) yields 180–240 W. To achieve the targeted 10 W of electric power from walking, therefore, a relatively large proportion of this energy must be converted. Thus, the design goal becomes one of maximizing energy conversion efficiency.

Energy Transduction Alternatives

The alternatives for converting mechanical energy to useful electric energy are few. A typical generator, which relies on electromagnetic coupling between a magnetic field and conductor moving relative to one another, is the most common approach. Although one can postulate several configurations of such a generator, rotary and rectilinear, inevitably these require high relative velocity between the armature and the stator. Efficient conversion of the relatively slow movement involved in walking was concluded to be unrealistic.

An alternate approach, preferred in the current application, was the use of piezoelectric material for directly converting mechanical energy to electrical energy. Piezoelectricity arises from the shift in electric polarization produced by mechanical strain in certain crystals, the polarization being proportional to the amount of strain. A certain group of crystalline solid materials is known to exhibit the piezoelectric effect, wherein an applied electrical field elicits a mechanical strain. Conversely, when these materials are mechanically stressed, they produce an electric charge. These materials are most commonly applied to sensors and actuators, at relatively high frequency of operation, and are not typically used for generating significant power. Accordingly, the literature is rather sparse regarding the theory or specifications relevant to the latter application. Therefore, the current feasibility study required additional investigation into the power generation capabilities of these materials.

Comparison of Piezoelectric Materials

Certain asymmetric mineral crystals, such as quartz, tourmaline, and Rochelle salts exhibit the piezoelectric effect to a moderate degree. Other ceramic materials such as lead zirconate titanate (PZT), barium titanate, and potassium dihydrogen phosphate exhibit this effect to a more pronounced degree. The most common piezoelectric materials available are polycrystalline compositions such as PZT, barium titanate, and lead metaniobate. Recently, ferroelectric polymer formulations of polyvinylidene fluoride and polyvinylidene difluoride (PVDF) have been developed that demonstrate piezoelectric properties.[7] The key properties of these materials, summarized in **Table 1**, involve their electrical characteristics, such as resistivity (P) and permittivity (ε); their mechanical properties, such as elastic modulus (E), strength (S), and density; and their piezoelectric activity, characterized by a piezoelectric strain constant (d) and stress constant (g).

Compared to the ceramic materials, the polymer materials are far more flexible, tough, and lightweight. These properties, in addition to their ease of fabrication into complicated shapes, would appear to make them ideal for the current application. These desirable properties, however, are overshadowed by their poor electromechanical coupling. Their low quality factor (Q), which makes them desirable for broadband applications, consequently results in relatively low electromechanical conversion and renders them less attractive for power generation.

Table 1. Comparison of Various Piezo Materials

	PVDF[4]		VF_2/VF_3[5]		PZT[9–11]		$BaTiO_3$[11]	Quartz
	31	33	31	33	31	33		
Density (10^3 kg/m^3)	1.78		1.82		7.5		5.7	2.65
Volume resistivity (ohm-m)	1.5e13		> 1e14		≈ 10e13		≈ 10e13	10e12
Permittivity [(F/m)]	106–113e-12 @ 1–10 kHz		65–75 e-12		10.5e-9		15e-9	39e-12
Piezo strain constant, d [(pC/m^2)/(N/m^2)]	23	−33	11	36e	100–274	153–593	78	2.3
Piezo stress constant, g [(V/m)/(N/m^2)]	216e-3	−339 e-3	162 e-3	529 e-3	9.1–16e-3	19–41e-3	5 e-5	58e-3
Young's modulus (Pa)	2-4e9		3-5e9	3.3e9	SC: 6.1–9.3e10 OC: 9.7–14.4e10	SC: 4.4–7.2e10 OC: 10.6–12.7e10		8e10
Strength (yield, ultimate) [MPa]	S_{yt}: 45–55 S_{uc}: 140–350		S_{yt}: 20–30 S_{ut}: 35		S_{ut}: 20–60 S_{uc}: 520		S_u: 52–59	S_{ut}: 27 S_{uc}: 1447
Maximum operating stress (MPa)	60–160		20e6		T: 20e6[8] 17–34e6[9] C: 400e6[8] 34e5[10] 66–126e6[12]		S_f: 21–24	—
k (rated k^2)	5% (0.25%) @ 10 Hz 12% (1.4%) @ 1 kHz	14% (2%) @ 100 Hz 29% (8.4%) @ 100 MHz	20% (4%)	29% (8.4%)	30–39	62–75	33: 15–21 31: 2.5–3.6	33: 10 31: 2
Maximum field [V/μm] operating (breakdown)	30 (80)		NA		0.5		NA	

PVDF, polyvinylidene difluoride; PZT, lead zirconate titanate; OC, open circuit; SC, short circuit; NA, not applicable; T, tensile; C, compressive.

Energy Conversion

Considering a piezoelectric element within the midsole of the shoe, interposed between the plantar surface and the outer sole, the resulting energy imparted from the foot will be distributed into three components, modeled as a generator, spring, and dashpot:

$$W_t = W_e + W_m + W_d \tag{2}$$

where W_t = total energy imparted by the foot, W_m = strain energy stored in the midsole,

$$W_m = \frac{1}{2}(\sigma e)\cdot V = \frac{1}{2}(Ee^2)\cdot V = \frac{1}{2}(\sigma^2/E)\cdot V, \tag{3}$$

W_e = electrical energy converted,

$$W_e = \frac{1}{2}Qv = \frac{1}{2}Cv^2 \tag{4}$$

$$= \frac{1}{2}(\epsilon g^2\sigma^2)\cdot V$$

and W_d = energy dissipated due to viscous losses/friction. Here σ is the stress and e is the strain in the piezo material, E is the elastic modulus, Q is the electrical charge, v is the voltage, and V is the volume of material. Assuming W_d to be negligible, this implies the following theoretic electromechanical conversion ratio:

$$\eta = \frac{\text{electrical energy withdrawn}}{\text{total energy in}} = \frac{\eta'}{\eta' + 1} \tag{5}$$

where

$$\eta' = \epsilon g^2 E = \frac{W_e}{W_m}. \tag{6}$$

The respective values for PVDF and PZT are presented in **Table 2.** Although all the materials display similar energy density characteristics, the polymer materials, because of their relatively low elastic modulus, require significantly more mechanical energy to accomplish the same electric conversion. This essentially eliminates them from consideration in the current application.

Note that η is not the same as efficiency because the mechanical energy stored in the transducer is recoverable upon toe-off and heel-off. However, the target energy output dictates that the transducer must be capable of converting greater than 50% of the imparted energy of the foot to maintain the midsole displacement within the desired limits.

The geometric requirements for the piezoelectric element can be evaluated by considering the energy density (joules/m^3) of the piezo crystal (derived in the Appendix):

$$\frac{W_e}{V} = \frac{1}{2}\epsilon g^2\sigma^2 = A_e\sigma^2 \tag{7}$$

Here, the parameter A_e is introduced to indicate the "energy density gain" of the material. Thus, the energy density depends proportionally on energy density gain, a property of the material, and the square of the applied stress. Comparison of various piezo materials under axial and lateral modes of strain reveal that A_e is similar for most materials (**Table 2**). Therefore, size-effectiveness is governed by the maximal

Table 2. Derived Energy Conversion Properties

	PVDF[6]		VF_2/VF_3[6]		PZT[6–11]		$BaTiO_3$[11]	Quartz
	Lateral(31)	Axial(33)	31	33	31	33		
Energy density grain, A_e [(J/Pa²)/m³] (1/2 · g²)	2.47e-12	6.09e-12	9.18e-13	9.79e-12	2.1e-12	5.3e-12	19e-18	6.6e-14
Maximum energy density [kJ/m³] ($A_e\sigma^2$)	25	61*	0.4	4	9.6–33	24–83	—	0.3
Electromechanical conversion theoretical {$2A_eE$} (rated k^2)	1–2% (1.4)	2.5–5% (3)	0.6–1% (4)	6% 8.4	25–29% (10–15%)	47–56% (43–56%)	(0.6–4%)	1%

* Maximum measured energy density: 200 kJ/m³ at ultimate stress.[6]

sustainable stress of the material, which in turn is governed by piezoelectric degradation, or depolarization, which occurs with repeated loading. This stress level depends upon loading rate and operating conditions, rendering it difficult to determine definitively. Consequently, these specifications are not commonly reported for most commercially available piezoelectric materials. It is in most cases considerably lower than the mechanical strength (i.e., yield, or ultimate strength) of the material. For example, energy densities as high as 210 kJ/m³ have been reported for PVDF,[7] but these have been obtained under single-action conditions wherein the material was strained to failure. Subjecting this material to cyclic loading, as would be expected in the foot generator, one could expect to extract 25–61 kJ/m³. Thus, the total volume required may be excessive. The comparatively higher energy density of PZT (as high as 83 kJ/m³) listed in **Table 2** is attributed to the higher estimated operating stress level.

The loading mode is an additional design consideration. The current design uses uniaxial loading because this results in the maximum energy conversion as compared to lateral, bending, shear, and hydrostatic loading. Because PZT ceramics can withstand much higher stress in compression than tension, due to their relatively limited fracture toughness, the current design uses a compressive column. For each heel/toe strike to generate 2.05 J of energy, in light of the maximum recommended stress of approximately 10×10^3 psi,[12] this will require 8.5×10^{-7} m³ of PZT material.

Shoe Design

A shoe generator was designed (**Figure 3**) based on the above calculations, featuring a double acting mode of operation. Two longitudinal barrels house cylindrical PZT piezoelectric stacks (200 mm length, 16 mm diameter), which are actuated by hydraulic amplifiers at each end. The master pistons of the hydraulic amplifiers are located approximately beneath the tarsometatarsal joint of the forefoot and calcaneal region of the heel, and are angled to align with the maximum reaction forces of the heel-strike (−7.8° to normal) and toe-off (9.5°). The total mechanical advantage of the system is 35:1. Thus, 10 mm of displacement of either the heel or the forefoot translates into 0.29 mm displacement of the stack.

The applied load, which appears at a relatively low frequency (**Figure 2**), can be subdivided into any number of smaller "packets" to the transducer. For this purpose, a hydraulic oscillator is incorporated into the master piston to convert the constant stroke to a higher frequency (5×) pulsed excitation of the stack. Although the same total energy will be imparted to the midsole, the increase in frequency has three beneficial effects: a reduction of the overall volume of the transducer, an increase in the maximum allowable stress, and an increase in electromechanical efficiency.

Power Conditioning

Power delivery is determined by the relative output of the remote shoe generator to the power drawn by the electrical load. During sedentary operation, power will be transmitted along the conventional path from the external battery source to the internal battery charger of the implanted device. When the shoe generator is activated, its energy will be directed via diode switching onto the power bus. In this condition its output will partially supply the power demands of the VAD, effectively extending the running time of the battery. Should the athletic effort on the shoe generator result in a sufficiently large power output, it is conceivable that the power demands of the VAD will be satisfied as the external battery is recharged. For the current studies, a simplified power conditioning circuit consisting of a bridge rectifier, a buffer capacitor, and a load resistor was configured as a basic charge pump. The capacitor can be connected directly to the external battery by an additional diode to provide a charging current dependent on the power delivery from the

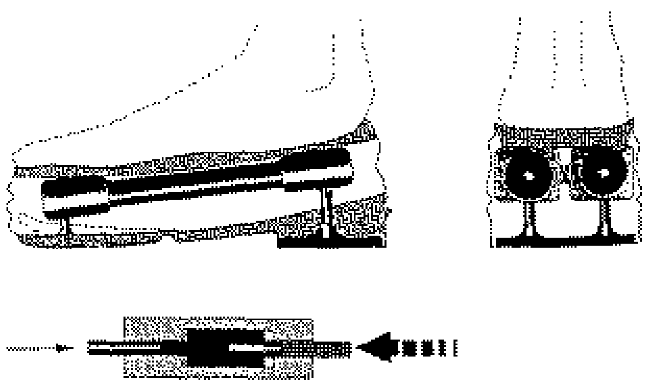

Figure 3. Schematic of shoe generator within midsole. Piezoelectric stack is actuated by hydraulic amplifiers coupled to heel and toe.

generator. For the current studies, a digital coulombmeter (Model 32504; Central Scientific Co, Franklin Park, IL) was used to measure the developed charge on the buffer capacitor.

The potential developed by the piezo stack is proportional to transducer thickness. For a given volume, a tall, slender stack will deliver a higher voltage than one with a smaller aspect ratio. By electrically coupling the piezo elements in parallel, any problem caused by high voltage would be obviated. In the current application, high voltage was not a concern because the electric load was large enough to prevent excessive charge from building up within the crystals.

Validation Experiments

Preliminary experiments on the PZT ceramic were conducted using a test cell to determine the effect of loading and electrical impedance matching under simulated walking dynamics. The apparatus consisted of a mechanical lever actuated by an electric solenoid (Densitron, Torrence, CA) powered by a function generator to simulate various foot loading conditions. Power output from the PZT element was measured for different electric loads and frequencies.

A second set of experiments using a 1/17 scale model of the prototype midsole assembly, similar to **Figure 3**, was constructed to validate the design principles postulated earlier. This device consisted of a single cylindrical stack of 18 PZT ceramic slugs (PZT-5A, 0.31 inch diameter, 0.245 inch thick; Morgan Matroc, Bedford, OH) within an insulated stainless steel barrel. Mechanical advantage ($r = 7.6$) was obtained by a simplified hydraulic amplifier without oscillation.

Power output was measured for four subjects (52 kg woman, 58 kg woman, 66 kg man, and 75 kg man) for three gait patterns: 1) walking at preferred cadence (approximately 1 step/sec) with a natural heel contact toe-off pattern; 2) flatfooted (both heel and toe contacting the ground simultaneously); and 3) jogging in place (at approximately 2 steps/sec). The subjects walked in shod feet with the generator attached to the right shoe sole and a passive orthosis attached to the left.

Results

The effective power transfer in the test cell was found to depend upon the stress application rate and the load impedance. Because the power developed on the piezoelectric element was available at very high open circuit voltages (e.g., 1000 V) across a relatively small capacitance (68 pF), at a low frequency, it was difficult to match the impedance using a resistive load. Several capacitive loads were tested to determine the optimal match. An inverse relationship between capacitance and developed voltage was observed. Larger capacitances (e.g., 10 μF) facilitated charge accumulation, at the expense of power transfer. The coulombmeter provided the most convenient means to measure charge. The rate of loading was found to affect the peak voltage directly, whereas it appeared not noticeably to affect the total charge developed.

The power generated through the walking experiments are listed in **Table 3** along with the estimated theoretic output. These represent the greatest outputs observed, which were obtained with a matched electric load. Walking experiments produced an average power of 5.7 ± 2.2 mW/kg body weight, with jogging providing a higher average power level of 23.6 ± 11.6 mW/kg. These results suggest that the power generation would reach approximately 6.2 W for the 75 kg subject operating a pair of full size midsole generators.

The foot contact pattern appeared minimally to influence the power generation. Whether the subject walked with a typical heel–toe pattern versus flat-footed, the power generation was not dramatically affected. However, further analysis needs to be performed to characterize the efficiency of power transfer. The point of contact with the foot was found to influence the output primarily through its effect on the gait pattern. When optimally aligned, the gait was most regular and electric output was maximized.

Discussion

The human body is simultaneously a consumer of energy and a power plant for interconverting energy for its continued operation. As remarked by Iles, "Another notable case of efficiency in nature . . . [is] the conversion by the animal frame of fuel-values into mechanical work."[13] The heart and other tissues receive power from this efficient delivery system. Only when we seek artificial substitutes for our carbon based organs must we also pursue additional energy to support life.

The idea of harvesting autologous power from the contractile forces of skeletal muscle is a natural approach to the problem of limited supply of external power. This has prompted investigators to consider applying skeletal muscle for direct actuation of ventricular assist devices. To achieve continuous skeletal muscle power output,[14-17] however, requires several weeks of muscle conditioning before continuous power can be extracted. A much less ambitious and less invasive approach to harnessing power from skeletal muscle is the basis of the current study. Rather than interpose a transducer internally into the force train of the skeletal muscle, the current study contemplates extracting power externally. A disadvantage of such an approach is that the power would be available only while the patient is ambulating, and only when the special shoes are donned. Although sedentary patients will not benefit from this on-board source of power, neither will they need to be untethered from external power.

The theoretic power estimates exceeded the recorded values for moderate gait in these preliminary experiments. The results of the test cell experiments would indicate that the discrepancy is the result of the uncertainties associated with the rated piezo constants, and with the need for proper electrical impedance matching. The piezo characteristics are highly dependent on experimental conditions, such as frequency, electrical impedance, and loading rate. The permittivity is also likely to increase as the limiting stress is approached. Mechanical constants, such as the elastic modulus, are also variable. For example, as charge is drained off, further deformation in the piezo will occur. The Young's modulus thus depends on the electric load. The shortages could also be dramatically reduced by more advanced power conditioning. Although power levels measured for the simulated jogging conditions were much higher, these levels were beyond the operating range of the device, and could not be sustained without degrading the piezo material.

Table 3. Measured Versus Theoretical Power Output From 1/17 Scale Model Midsole Generator

Subject No.	Weight (kg)	Gender	Estimated Theoretical Power (mw)	Measured Power		
				Flatfoot	Heel-Toe	Simulated Jog
1	52	Women	421	306	256	676
2	58	Women	521	156	361	900
3	66	Man	664	342	225	2500
4	75	Man	942	625	676	2100

Strain energy diverted to the generator on release of sole loading would theoretically increase the energy expenditure of the patient because the shoe generator would emulate energy absorbing sole characteristics. As a best case, the energy converted in the shoe generator is analogous to energy dissipated in a running shoe,[5] wherein energy dissipation is diminutive with respect to the energetics of the muscles involved in ambulation. As shown by Bosco and Rusko,[18] shoes with energy absorbing or damping soles tend to increase the oxygen consumption of treadmill runners. In the limit, the additional perceived work introduced by the elasticity of the shoe may be likened to walking on a (stiff) spring mattress, whereas the nonelastic component may be (roughly) compared to walking on loose sand, or a shallow stair climbing exercise.

In a conventional shoe, viscoelastic characteristics of the sole material are known to affect the distribution and levels of strain and dissipated energy during the application and subsequent release of foot loading. It has been estimated[5] that work done on the sole of a running shoe having relatively low stiffness and damping characteristics can be as high as 18 J per step. For the same sole, the elastic work recovered is approximately 13 J, whereas the remaining 5 J is dissipated.

Because the power levels involved in walking and running differ significantly, there is a choice whether to size the generator to accommodate the maximum expected (running) power level versus a lower level baseline. The former would result in an excessively oversized shoe for normal walking. It would therefore be more convenient for the subject to change shoes before running or jogging. Therefore, when patients anticipate the need to increase their activity level, and desire to do so without affecting their battery life, they may choose the appropriate shoe. Besides the added potential for muscle fatigue, additional factors that would influence patient acceptance of this device would include stability, comfort, noise, weight, and aesthetics. These factors should be addressed in the production of the next generation prototype.

Two main criteria for mechanical coupling from the foot to the transducer are force transmission and loading frequency. One would expect the force to depend on compliance matching of leg to shoe and foundation.[19] In fact, the forces prove to be rather insensitive to compliance. Human feedback appears to regulate the reaction force to achieve the desired acceleration. Therefore, the force coupling is governed primarily by the requirements for maximal energy extraction by the transducer. The latter requires that the developed stress approximate the maximum operating strength, and that the frequency of loading also be maximized, within limits.

To develop the required stress, the foot force must be concentrated over a relatively small area. Direct coupling is the most desirable approach for design simplicity. Unfortunately, this would require an extremely small cross-sectional area of transducer to be stressed. The consequence would be an infeasible transducer length. This requirement also precludes the use of a distributed transducer array in the midsole. Thus, a concentrated approach requires additional mechanical advantage between the point of contact of the sole and the transducer. In this case, it is logical to use two contact points for mechanical coupling. An additional advantage of such a concentrated transducer is that a single transducer can be used for both the heel and toe, minimizing dwell time. This design also proves to be relatively insensitive to variations in the foot contacting pattern (e.g., "heel striking" versus "toe striking").[20]

Although the weight of the actual transducers is not very large (640 g for the prototype; 40 g for the scaled model), the related chassis, linkages, and couplings can accrue significant weight. Although the transducers could be located remotely to the shoe (e.g., on a belt pack) with hydraulic lines coupling them to a master piston in the midsole, an integrated design is more desirable. The hydraulic amplifiers used in the current design allow a dramatic reduction in the axial force exerted on the stack housing, thus reducing the required stiffness (and thus weight) of the chassis, and limiting the possibility of jamming within the mechanical linkage.

Inasmuch as the developed power is directly proportional to the frequency, any multiplication of frequency would directly benefit an improved energy/size ratio. In addition, the piezo conversion characteristics are known to be sensitive to operating frequency. Very low frequency operation (1 Hz) could impair energy conversion by as much as 50% from optimum. Mechanical oscillation could be accomplished in a number of ways, including a ratchet with overrunning clutch drive, or with hydraulic valving as in the current design. An alternative to the hydraulic pulser is a mechanical resonator. A resonant system would be excited by impact, and would "ring" at the resonant frequency determined by the mass and elasticity of the system. This is not a very efficient form of coupling, however, because it transfers only a limited bandwidth of the incipient energy. The mechanical energy could be buffered, for example in a spring motor or flywheel, yet these are limited both by size and efficiency loss.

In spite of the discrepancy observed in the current prototype between theoretic and measured power output, the results of this feasibility study motivate further development of an improved full scale prototype. This will allow experiments to be conducted to evaluate the extent to which the shoe generator design affects the kinematics of gait and the associated energy expenditure (e.g., oxygen cost). Harmonic analysis of torso acceleration, which has been used as an index to assess relative "smoothness" of walking after limb or joint immobilization,[21] will be applied for this purpose.

Because artificial prosthetic organs lack the elegance of design displayed by the biologic systems they supplant, so too does the device described herein lack the sophistication and power conversion efficiency of the biologic generator it is attempting to harness. This brings to mind the observation of Petroski that, "We are all engineers of sorts, for we all have the principles of machines and structure in our bones. . . . We may wonder if human evolution may not have been the greatest engineering feat of all time."[22]

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Appendix: Derivation of Electric Energy Density

The electric energy stored on internal capacitance is given by the familiar formula:

$$W_e = \frac{1}{2} Qv = \frac{1}{2} Cv^2 \tag{8}$$

where the capacitance of the piezo crystal is given by:

$$C = \epsilon \frac{a}{t}. \tag{9}$$

Here, ϵ is the permittivity, a is the electrode area, and t is the plate separation. The potential is found by the product of the electric field density ($\bar{E}$) and plate separation:

$$v = \bar{E}t \tag{10}$$

where

$$\bar{E} = g\sigma \tag{11}$$

and g is the characteristic piezo stress constant. Substituting equations (9), (10), and (11) into equation (8) delivers:

$$W_e = \frac{1}{2} (\epsilon g^2 \sigma^2) at \tag{12}$$

which is equivalent to equation (7) in the text.