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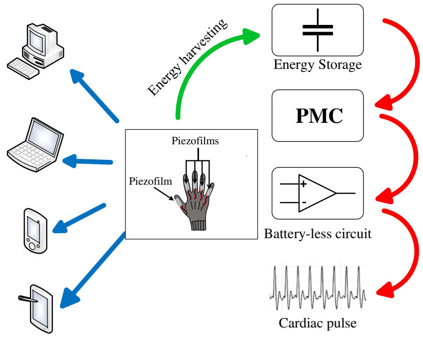
Battery-Less Power Management Circuit Powered by a Wearable Piezoelectric Energy Harvester



The wearable system proposed herein can harvest and manage the energy from daily activities like using a laptop, computer, cellular phone, or tablet. The energy can be used to supply low-power circuits; in our case, the proposed system is tested by detecting the cardiac pulse for 20 s with no batteries, in a comfortable and user-friendly way.

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Battery-Less Power Management Circuit Powered by a Wearable Piezoelectric Energy Harvester

Q1 *Omar Mendez-Lira, Enrique Mario Spinelli, and Rafael Gonzalez-Landaeta**

A system that aims to manage the energy harvested from the finger movement using a glove-type garment is proposed. For this, five polyvinylidene difluoride (PVDF) film-type sensors are mounted on different interphalangeal joints of the fingers. A 68 µF/25 V ceramic capacitor is used as a storage device. A battery-less power management circuit (PMC) is proposed to isolate the capacitor from the load to reduce the current consumption while charging. When the voltage across the capacitor reaches a threshold voltage, the user decides when to transfer the energy to the electronic load. To test the proposed system, the harvested and conditioned energy is used to power both the PMC and a battery-less cardiac pulse detection circuit, achieving an autonomy of 20 s. This time depends on the voltage stored in the capacitor and the current consumption of the load. Using the proposed wearable system, it is possible to store up to 700 µJ in the capacitor when a subject manipulates a computer or a cellular phone for 10 min. The proposed system is not limited to the activities presented herein because it can harvest the biomechanical energy from any activity that involves finger movements.

1. Introduction

Q2 The need for providing a continuous amount of energy is one of the main concerns in wearable devices.[1,2] The limited power source (typically rechargeable batteries) has a direct impact on the size, weight, lifetime, and maintenance of the device.[3] To tackle this, some authors have proposed different approaches like using energy harvesting (EH) for wearable devices to reduce the use of batteries or even replace them.[4]

EH systems transform nonelectric energy from different sources into electric energy to be stored and used later. These sources can be found in the environment or in the human body. Energy from the environment can be motion, thermal gradients, and RF [4,5] and the energy from the human body can be in the form of heat, biochemical and biomechanical, and it is more predictable.[4] Biomechanical energy is attractive for wearable systems considering the dynamic nature of the human body. However, most of the sensors developed to harvest this kind of energy are more efficient at high frequencies and low displacements.[6] However, in daily life, the dynamic of the human body is low frequency, random, and with large movements.[7,8]

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EH from the human motion captures the biomechanical energy from different activities that involve center of mass motion, joints motion, foot strike, and limb swing motion.[9] More recently, efforts have been made to harvest energy, while the subject is doing daily activities such as using a computer, a smartphone, or even eating.[7,10–13] In that sense, designing a wearable energy harvester becomes more challenging due to the strict physical constraints of wearable systems in terms of size, weight, form, and, in the case of clothing, mechanical flexibility.

Piezoelectric EH has been demonstrated to be an excellent technique for self-powered wearable systems.[14] Commonly, polymer-based, flexible materials, such as PVDF, rather than rigid and brittle ceramics, such as lead zirconate titanate (PZT), are used.[15] However, as the electrical energy obtained with PVDF sensors is low, intermittent operation with low duty cycle becomes mandatory.[16] This problem has been tackled using power management circuits (PMCs) to accumulate energy in a storage device and then deliver it to the load. In that case, the load not only receives a regulated voltage intermittently but must deal with under-voltage conditions and avoids problems such as the lock-up phenomenon described in other works.[17,18]

In this article, a self-powered glove-type wearable system is proposed to harvest and manage the energy from finger motion. To convert the biomechanical energy into electrical, five piezoelectric films sensors are used. The electrical energy is first conditioned and stored in a capacitor and then is delivered to the load. For this, a battery-less PMC is proposed to isolate the capacitor from the load during the charging phase, whereas the current consumption of the circuit is reduced. Once the level of energy reaches the requirements of a low-dropout linear voltage regulator (VR), the user decides when to transfer the stored energy to the load. In this work, the proposed system is tested by power supplying a cardiac pulse detection circuit. A similar piezoelectric glove has been already proposed by Psoma et al.,[10] where the

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harvested energy was used to charge a nickel–metal hydride (NiMH) battery from the impact of the fingers over a surface using piezoelectric sensors located at the fingertips. This approach can be obstructive and uncomfortable when using a keyboard, computer mouse, a cellular phone, or a tablet. The system herein proposed is unobtrusive and user friendly.

The manuscript is organized as follows. Section 2 describes the EH principle each stage of the energy-conditioning system. Section 3 explains the work principle of the proposed battery-less PMC and analyzes the effect of the circuit losses. Section 4 describes the battery-less circuit used to detect the cardiac pulse. Section 5 describes the setups and the measurements to characterize the energy harvesters and the energy-conditioning circuit. Section 6 presents and discusses the results and Section 7 draws the main conclusions.

2. EH and Conditioning

The proposed system harvests the energy produced by the finger motion. For this, five PVDF film-type piezoelectric sensors (piezofilms) are mounted in a glove-type garment. As the subject puts it on, the piezofilms match different interphalangeal joints of the fingers to generate higher voltages, while the subject moves his/her fingers.[19] The sensors are placed on the posterior-proximal interphalangeal joint of the thumb and the posterior-proximal interphalangeal junctions of each of the other four fingers. With this distribution, the sensors do not generate voltage from impacts but rather from the movements of each finger.

The movements of the fingers are low frequency, high amplitude, and random, so the generated voltage from the harvesting sensors is not suitable to power an electronic circuit continuously. **Figure 1** shows the proposed circuit for conditioning and managing the energy harvested by the piezoelectric sensors.

The generated voltage is full-wave rectified by a rectifier bridge implemented with diodes with low-forward voltage and low-leakage currents, which are the main sources of power loss.[20] An over-voltage protection is placed to prevent the storage capacitor and the VR from being damaged when the energy is still being harvested and the system does not deliver the energy to the load. For this, a Zener diode, D_Z, with a voltage, V_Z, is used.

A diode, D, is placed between D_Z and the capacitor to reduce the current losses because of the leakage currents of D_Z and the rectifier bridge. The combination of D_Z and D makes the maximum voltage across C, V_{C_max}, not exceed the maximum input voltage of VR, V_{IN_max}. A capacitor C is used as the energy-storage device; its value should not be so high to shorten the charging time, nor so low to maintain the stored voltage for longer; this trade-off will be discussed in the following subsections. C must have a high insulating resistance to maintain the harvested energy longer when no energy is harvested. Although capacitors are not recommended as storage devices when energy is harvested from limb motions (walking or typing),[21] this work shows that good management of the harvested energy makes the capacitor a suitable option for battery-less applications.

3. Power Management Circuit

The proposed PMC is a battery-less circuit that alternates between an energy-storage phase and an energy-delivery phase. During the former, C is isolated from the load while being charged using the energy harvested from the fingers. During the latter, the stored energy is transferred to the load, making C discharge gradually. The energy-delivery phase lasts until the stored voltage reaches a low threshold voltage, V_L. In our case, the user decides when to power the load, but first, C must reach a high threshold voltage V_H. Both threshold voltages must be adjusted between the allowable voltage range of the VR, that is, $V_{IN\_min} \leq V_L < V_H \leq V_{IN\_max}$, $V_{IN\_min} = V_{CC} + V_{DO}$, where V_{CC} is the regulated output voltage, V_{DO} is the dropout voltage, and V_{IN_in} is the minimum input voltage at which VR begins to regulate, respectively.

A low-dropout and linear VR with a built-in ON/OFF circuit is used to deliver a steady and regulated voltage to the load. Although linear regulators are less efficient than switching ones, the former are simpler to use, low cost, have low noise, and have no AC switching losses.[22] In the system proposed here, when VR is off, C is isolated from the load; when VR is on, the stored voltage in C is transferred to the load. The ON/OFF control is achieved by two voltage dividers (R_1-R_2 and R_3-R_4), an operational amplifier (OA1) configured as a voltage comparator, and a

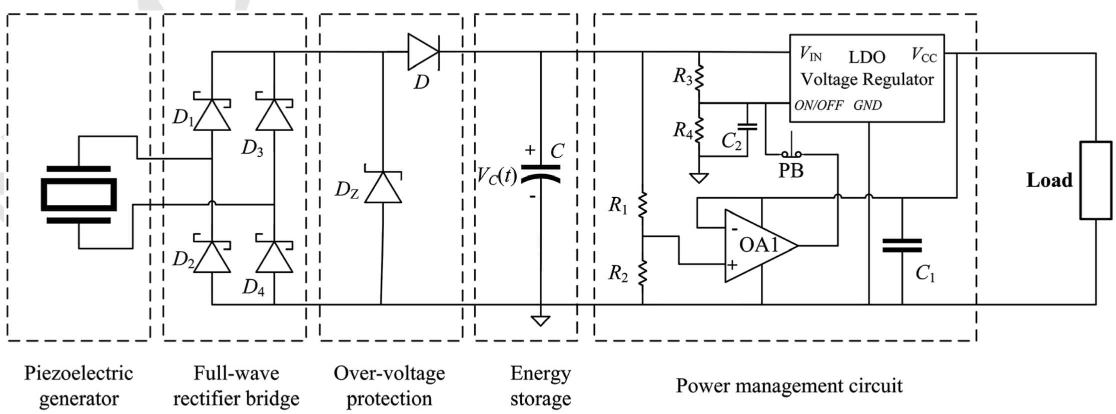

Figure 1. Proposed EH, conditioning, and PMC.

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normally closed push button (PB), which opens when pressed and closes when released.

In EH systems, PMCs work by switching or triggering between energy-storage and energy-delivery phases[22,23] but under certain conditions, usually threshold voltages and/or load requirements,[18] so the load is powered intermittently. In this proposal, a different approach is used. First, considering C as initially discharged, $V_C(0) = V_{IN} = 0$ V. When VR is off, the full-wave rectified voltage charges C, causing $V_C(t)$ and V_{IN} to increase. As the output of VR is 0 V, OA1 is off. In this case, the voltage divider formed by $R_1 - R_2$ is not involved with the startup or shutdown of the system. The input current coming from C finds a ground path through R_3, PB, and the low output resistance of OA1, making $V_{ON/OFF} = 0$ V, maintaining VR off. Although $V_C(t) \geq V_{IN\_min}$, VR is still off until PB is pressed. The instant when PB is pressed, the ground path of the input current is through R_3 and R_4, and now $V_{ON/OFF}$ is

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$$V_{ON/OFF} = V_C \frac{R_4}{R_3 + R_4} \tag{1}$$

Knowing the value of V_{ON} (minimum voltage of $V_{ON/OFF}$ to activate VR), the values of R_3 and R_4 can be estimated for a desired $V_C(t)$ ($= V_H$) that turns on VR when PB is pressed. Assuming a high value of R_3 to reduce the current consumption, R_4 can be obtained by

$$R_4 = \frac{V_{ON}}{V_C - V_{ON}} R_3 \tag{2}$$

Once PB is released, if $V_C(t) \geq V_H$, the VR output changes from 0 V to V_{CC} due to the output of OA1, and the energy-delivery phase begins. OA1 is configured as a voltage comparator; whenever $V^+ > V^-$, $V_{ON/OFF}$ changes to V_{CC} and VR is on until $V_C(t) = V_L$. At this point, R_1 and R_2 make $V^+ \leq V^-$, and the OA1 output and $V_{ON/OFF}$ change from V_{CC} to 0 V, turning off the VR and stopping the energy delivery to the load. If PB is not released, OA1 cannot control the state (on/off) of VR.

Assuming R_2 to be very high to reduce the current consumption, for a given V_L ($= V_{IN\_min}$), R_1 can be estimated by

$$R_1 = \frac{V_L - V^+}{V^+} R_2 \tag{3}$$

When VR is off, the current consumption is considerably reduced, lowering the discharging rate of C. In Figure 1, C_1 reduces the ripple of the supply voltage of OA1 and C_2 reduces the ripple of V_{CC} when PB is pressed and released.

3.1. Energy Storage

For the analysis presented here, we have considered the equivalent circuit shown in **Figure 2**a for each piezoelectric sensor.[16]

In our case, the output signal of the piezoelectric sensors is full-wave rectified (Figure 2b) instead of using a voltage doubler. Considering just one piezoelectric sensor in Figure 2b, C_S, which is the output capacitance of the sensor, is discharged/recharged in an opposite way each semicycle, and this charge is accumulated in C at each semicycle of the input signal. Considering the

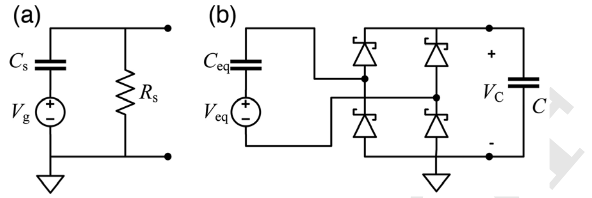

Figure 2. A) Equivalent circuit of a single piezoelectric sensor and B) equivalent circuit when the five piezofilms are electrically interconnected in series and in parallel to charge C, with V_{eq} full-wave rectified.

electric charge conservation law, the charge stored in C (Q_C) at a certain k-semiperiod, $k > 0$, is

$$Q_C(k) = Q_{Cs}(k) + Q_C(k-1) + Q_{Cs}(k-1) \tag{4}$$

$$CV_C(k) = C_S(V_p - V_C(k)) + CV_C(k-1) + C_S(V_p - V_C(k-1)) \tag{5}$$

Rearranging

$$V_C(k) = 2V_p\left(\frac{C_s}{C_s + C}\right) + V_C(k-1)\left(\frac{C - C_s}{C_s + C}\right) \tag{6}$$

Solving the discrete difference in Equation (6), considering $V_C(0) = 0$, $V_C(1) = V_p C_s/(C_s + C)$, and the forward voltage of the rectifier diodes (V_γ), the voltage across C as a function of k-semiperiod, results in

$$V_C(k) = (V_p - 2V_\gamma)\left[1 - \frac{C}{C - C_s}\left(\frac{C - C_s}{C_s + C}\right)^k\right]; k \geq 1 \tag{7}$$

where V_p is the peak amplitude of the input voltage. Using a full-wave rectifier, C charges twice per each cycle of the input signal, so, $k = 2\,t/T = 2tf$, f being the frequency of the input signal and t time. Now, the voltage across C as function of time and frequency is

$$V_C(t,f) = (V_p - 2V_\gamma)\left[1 - \frac{C}{C - C_s}\left(\frac{C - C_s}{C + C_s}\right)^{2tf}\right] \tag{8}$$

Considering $V_C(0) \neq 0$ and $C >> C_s$, Equation (8) can be rewritten as

$$V_C(t,f) \approx V_P\left[1 - \left(1 - \frac{V_C(0)}{V_P}\right)(1 - 2C_s/C)^{2ft}\right] \tag{9}$$

and the time needed for C to reach a given voltage V_x can be estimated by

$$t \approx \frac{C}{4fC_s}\ln\left(\frac{(V_P - 2V_\gamma) - V_C(0)}{(V_P - 2V_\gamma) - V_x}\right) \tag{10}$$

Although a high value of C would maintain the stored voltage for longer, it also implies that it would take a longer time to reach V_H. Hence, from Equation (10), some scenarios can be considered to reach V_H faster: 1) using an initially charged C, 2) increasing C_s, 3) increasing V_p, and 4) increasing the frequency of the deformations. Scenario (1) can be achieved by making $V_C(0) > 0$ V

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once the energy-delivery phase concludes. Scenario (4) is quite difficult to control because of the randomness of the movements. Nevertheless, Scenarios (2) and (3) are more predictable because the value of C_s and V_p can be determined by interconnecting the piezoelectric sensors in series or in parallel. However, it is necessary to assess which is the best option, because the effects on C_s and V_p are opposites.

In our proposal, each piezofilm harvests energy from the movements of each finger independently, so the harvested energy varies from sensor to sensor. Figure 2b shows the equivalent circuit when the sensors are interconnected either in series or in parallel to charge C, where R_s of Figure 2a has been disregarded due to its very high value.

Considering five identical sensors, in series, the total open-circuit voltage is the sum of the voltages generated by each sensor ($V_{eq} = 5\ V_g$) and $C_{eq} = C_s/5$. In parallel, the total voltage is the same of one sensor ($V_{eq} = V_g$) and $C_{eq} = 5C_s$. Considering the peak amplitude (V_p) of the open-circuit voltage of each sensor, from Equation (9), considering $V_C(0) \neq 0$, $V_C(t,f)$ for each topology is defined by

Series

$$V_C(t,f) \approx (5V_p - 2V_\gamma)\left[1 - \left(1 - \frac{V_C(0)}{5V_P}\right)(1 - 2C_s/(5C))^{2ft}\right] \tag{11}$$

Parallel

$$V_C(t,f) \approx (V_p - 2V_\gamma)\left[1 - \left(1 - \frac{V_C(0)}{V_P}\right)(1 - 2(5C_s)/C)^{2ft}\right] \tag{12}$$

For $t \to \infty$, it is obvious the series topology achieves $V_C = 5\ V_p$, approximately; so, at first instance, it is the best choice if C needs to be charged to a higher voltage. However, in the system proposed here, the PMC and the load only need +2 V, and it is not necessary for $V_C(t)$ to reach 5 V_p; otherwise, a lot of power will be lost due to the difference between V_{IN} and V_{out} of VR. Preliminary tests show that opening and closing the hand make each sensor provides 37 V, approximately. From Equation (10), considering V_x between 2 V and 5 V, the parallel topology is the best option to make C reach V_H faster.

Once $V_C(t)$ is estimated, the energy storage in C (E_C) can be obtained by

$$E_C = \frac{1}{2}C(V_{Cmax}^2 - V_{Cmin}^2)(\text{Joules}) \tag{13}$$

where V_{Cmax} and V_{Cmin} are the maximum and minimum voltage across C, respectively.

3.2. Circuit Losses and Energy Delivery

Figure 3 shows the equivalent circuit of the system of Figure 1, where the current loads and the equivalent resistances seen from C are presented. From this circuit, it is possible to estimate the discharge rate when the load is isolated from C and when the stored voltage is delivered to the load.

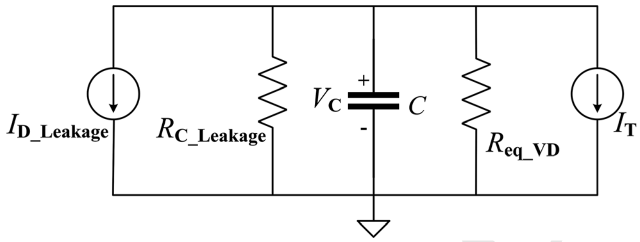

Figure 3. Equivalent circuit of the system of Figure 1 that shows the current loads and the equivalent resistances seen from C.

From Figure 3, $V_C(t)$ can be estimated by

$$V_C(t) = V_C(0)e^{-t/R_{eq}C} + IR_{eq}(e^{-t/R_{eq}C} - 1) \tag{14}$$

where $V_C(0)$ is the initial voltage across C, R_{eq} ($R_{C\_Leakage}||R_{eq\_VD}$) is the equivalent resistance, which depends on the insulating resistance of C, $R_{C\_Leakage}$, and the equivalent resistance of the two voltage dividers, R_{eq_VD}, and I is the current consumption of the entire circuit, which depends on the state of VR. For $t << R_{eq}C$, $V_C(t)$ is

$$V_C(t) \approx V_C(0) - (V_C(0) + IR_{eq})t/R_{eq}C \tag{15}$$

From Equation (15), the discharge rate of C can be approximated as linear. When VR is off, C discharges due to the current through the resistors R_1, R_2, R_3, R_4, and $I_{D\text{—}leakage}$ (the reverse current of D). If $I_{D\_Leakage}$ is very low, $V_C(t)$ depends mainly on $V_C(0) - V_C(0)t/R_{eq}C$. By making R_{eq} large, the discharge of C will be slower. When VR is on, the current drawn by VR and the load dominates, and $V_C(t)$ depends on $V_C(0) - It/C$, with $I \approx I_T \approx I_{VR\_ON} + I_{LOAD}$, where I_{VR_ON} is the current consumption when VR is on, and I_{LOAD} is the current consumption of the circuits of **Figure 4**. In this scenario, C discharges faster and the autonomy of the system depends on the time it takes C to reach V_L. The leakage currents of D_Z and the rectifier bridge are considered negligible because these diodes work at a voltage close to 0 V. This is because when D is reverse biased and D_Z is off, D acts as an open circuit, causing a large voltage drop between the cathode and the anode of D.

4. The Load

The load of the system of Figure 2 is the battery-less cardiac pulse detection circuit of Figure 4a, the voltage reference circuit of Figure 4b, and a Transflective Polarizer LCD Display with no backlight. In Figure 4a, when the cardiac sensor (piezoelectric sensor) detects the cardiac pulse, the output voltage of the sensor is amplified by a noninverting amplifier with gain $G = 1 + (R_B/R_A)$. R_x and the output capacitance of the cardiac sensor (C_{sensor}) define the time constant ($\tau = C_{sensor}R_x$) and the high-pass response of the system, which must be able to detect the cardiac pulse. A first-order passive low-pass filter formed by C_F and R_F reduces the noise bandwidth and the contribution of some electromagnetic interferences. The reference voltage of the circuit of Figure 4a is set to $V_{CC}/2$ using the system of Figure 4b. Both circuits are powered by a regulated voltage V_{CC} provided by VR. The current consumption of the circuits

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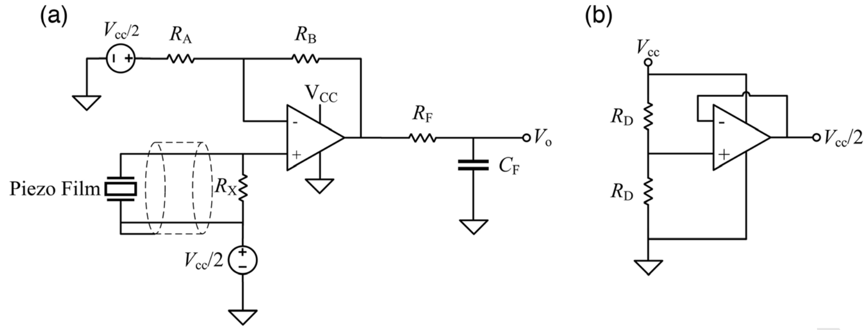

Figure 4. A) Cardiac pulse detection system and B) voltage reference circuit.

of Figure 4 depends on the quiescent current (I_Q) and the output current (I_{out}) of each amplifier, namely, $I_{LOAD} \approx 2(I_Q + I_{out})$. The circuit of Figure 4a draws more current because of the amplitude of the cardiac signal. The LCD indicates if the harvesting system is in the energy-storage phase (OFF) or in the energy-delivery phase (ON). The current consumption of the LCD has been neglected because it is quite lower than the op-amps.

5. Experimental Section

Table 1 shows the components used in each stage of the circuit shown in Figure 1 and their main characteristics, except the load.

To choose the optimal value of C, preliminary tests were made with capacitors of 68 µF, 100 µF, 150 µF, and 1 mF. To achieve a balance between both the energy-storage and energy-deliver phases, a 68 µF capacitor was chosen to implement the system of Figure 1 and conduct all the tests. The storage capacitor used here was ceramic, because the electrolytic ones, commonly used in this type of system,[24,25] had a higher self-discharge rate.[26] C can support up to 25 V, which was higher than V_{IN_max}. However, the combination of D_Z and D made V_C reach $V_z - V_F$, protecting C and VR. Maybe using a high-input voltage VR would avoid the use of D_z and D; however, using high-input voltage VRs implies that a higher voltage difference between input and output must be considered, causing further loss of power.[27]

To assess the effect of the electrical connection between all the sensors, real conditions were tested. For this, five LDT1-028 K from Measurement Specialties were mounted on a glove, which was used by a volunteer. The volunteer was then asked to close and open his/her hand following a 1 Hz and 2 Hz pattern during 20 s each. This experiment was conducted when the sensors were interconnected both in series and in parallel. The total output voltage of the five sensors was full-wave rectified to charge C; during this test, the circuit of Figure 2b was used, and $V_C(t)$ was measured using DAQ NI USB-6341 from National Instruments.

To estimate R_1 and R_4, we assumed $R_2 = R_3 = 100\ \text{M}\Omega$ to reduce the current consumption. $V_H = +5\ \text{V}$, $V_L = +2.25\ \text{V}$, and $V^+ = +1.99\ \text{V}$, which was enough to produce a change from 0 V to V_{CC} at the OA1 output. $V_{ON} \geq 0.63\ \text{V}$ and $V_{OFF} \leq 0.54\ \text{V}$. These values were estimated experimentally, measuring $V_{ON/OFF}$ with a 6½-digit Digital Multimeter 34461 A from Keysight when VR was turned on and off, respectively. From Equations (2) and

Table 1. Main characteristics of the components used to implement the system of Figure 1.

Stage	Item	Main Characteristics
Piezoelectric generator	LDT1-028 K	$C_s = 1.38$ nF
Full-wave rectifier bridge	Schottky Diodes BAR28	$V_F = 0.41$ V
		$I_R = 200$ nA @ 50 V (max)
Overvoltage protection	Zener Diode BZX55C6V2-TA	$V_Z = 6.3$ V (typ), 6.6 V (max)
		$I_R < 100$ nA @ 2 V
	Small-Signal Diode 1N4151TAP	$V_F = 0.8$ V (typ), 1 V (max).
		$I_R = 50$ nA @ 50 V (max)
Energy storage	Capacitor	68 µF, 25 V, Ceramic.
PMC	Voltage Regulator S-1313B20-M5T1U3	V_{IN}(max) = 6.0 V[a]
		$V_{CC} = 2$ V (1.0%)
		$V_{DO} = 0.23$ V (typ.)
	Operational Amplifier MCP6041T-I/OT	Current consumption:
		ON: 1.35 µA @ 3 V (no load)
		OFF: 100 nA @ 3 V (no load)
		$I_Q = 1$ µA (max.)
	Push Button	Normally closed
	R_1	13 MΩ, 1%
	R_2	100 MΩ, 5%
	R_3	100 MΩ, 5 %
	R_4	13 MΩ + 1.6 MΩ, 5 %
	C_1	100 nF, 25 V
	C_2	270 pF, 25 V

[a]Absolute maximum rating.

(3), $R_1 = 13.06\ \text{M}\Omega$ and $R_4 = 14.41\ \text{M}\Omega$. To implement the circuit, we used commercial values of 13 MΩ for R_1 and a series combination of 13 MΩ and 1.6 MΩ for R_4.

In Figure 4a, a shielded piezoelectric sensor SDT1-028 K ($C_s = 2.78$ nF) from Measurement Specialties was used as the cardiac sensor. The gain of the noninverting amplifier was 11 ($R_A = 1\ \text{M}\Omega$ and $R_B = 10\ \text{M}\Omega$), and the bandwidth, defined by R_F and C_F, was 10 Hz. $R_x = 30\ \text{M}\Omega$, so $\tau = 83.4$ ms. In Figure 4b, $R_D = 10\ \text{M}\Omega$ to reduce the current consumption. The op-amp used in both circuits was TLV522DGKR from

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Texas Instruments, which was a dual-nanopower amplifier, where each op-amp had a quiescent current $I_Q = 800$ nA (max), a wide supply range (from +1.7 V to +5 V), and a very high input impedance ($10^{13}\,\Omega \parallel 2.5$ pF). Both circuits were powered at +2.0 V provided by the low-dropout VR S-1313D22-M5T1U3 from ABLIC.

The measurement setup shown in **Figure 5** was used to conduct two tests, where C was initially charged to +5.15 V using a DC power supply E3631A from Keysight (not shown in the figure). In the first test, $V_C(t)$ was measured, whereas VR was off; the discharging rate was estimated when no energy was

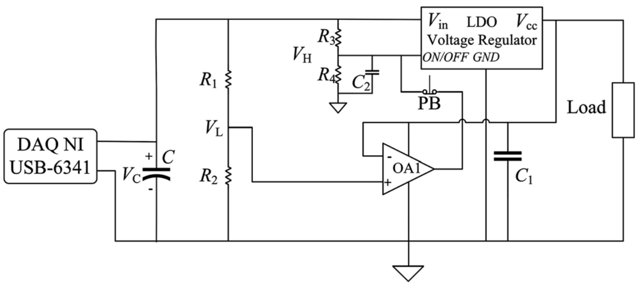

Figure 5. Experimental setup used to measure V_C when the VR was off and on.

transferred to the load. Then, $V_C(t)$ was measured when VR powered the PMC and the load. During this part of the test, the cardiac sensor was placed on the radial artery to detect the cardiac pulse using the energy stored in C. In the second test, once $V_C(t)$ reached V_L and VR was turned off, the volunteer opened and closed his/her hand to harvest energy, and the time needed for $V_C(t)$ to reach V_H was estimated. During both tests, $V_C(t)$ was measured by a DAQ USB6341 from National Instruments; the current consumption was estimated by $I_{avg} = C\Delta V_C/t$, and E_C was estimated by Equation (13).

Finally, the same volunteer, wearing the glove, used a computer (QWERTY keyboard and a mouse) and a cellular phone for 10 min each, without any restrictions. For this test, VR was off, the five piezofilms were interconnected in parallel, and all the stages of the system of Figure 1 were connected to C, $V_C(t)$ was measured, and E_C was estimated.

6. Results and Discussion

Figure 6 shows the open-circuit voltage (V_{eq}) of the five piezofilms connected in parallel (Figure 6a) and in series (Figure 6b), when the volunteer opened and closed his/her hand at 2 Hz using the glove. First, a nonperfect and nonsymmetric sinusoidal wave from finger movement is shown. Second, $V_{p\_parallel} \approx 5\ V_{p\_series}$, that is, 36 V

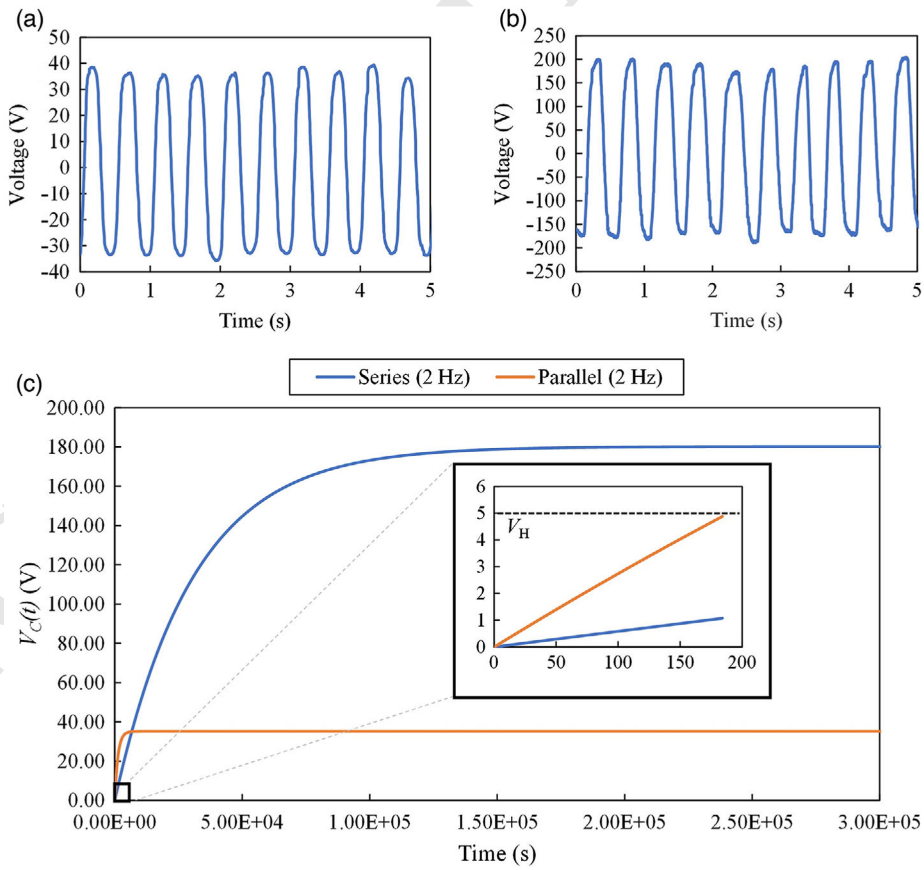

Figure 6. Open-circuit voltage of the five sensors when the subject opened and closed his hand at 2 Hz wearing the glove. A) Sensors interconnected in parallel and B) in series. C) Theoretical results of $V_C(t)$ during 300×10^3 s using both topologies.

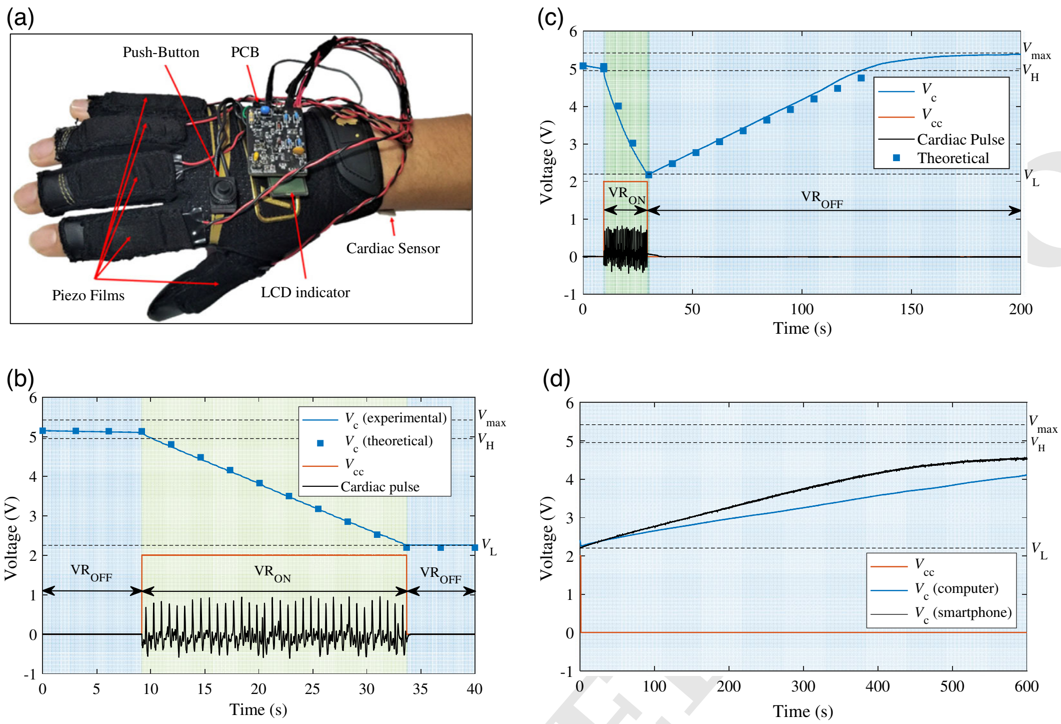

Figure 7. A) Photograph of the wearable system that harvests biomechanical energy to power a battery-less cardiac pulse detection circuit, B) voltage across *C* when VR is turned on and off, C) performance of the system when the energy is harvested as the subject opened and closed the hand at 2 Hz, approximately, and D) $V_C(t)$ (while VR was off) using the biomechanical energy of the fingers when the subject, wearing the glove, used a computer (keyboard and mouse) and cellular phone for 10 min each.

(parallel) and 181 V (series), approximately, making the series topology the best option to generate higher voltages. Figure 6c shows the theoretical results obtained from Equations (11a) and (11b), where $C = 68\,\mu F$, $C_s = 1.38$ nF, $V_\gamma = 0.4$ V, and $t = 300 \times 10^3$ s. $V_C(t)$ reaches almost 181 V for the series topology and 36 V for the parallel topology. However, these levels are never reached because of D_Z of Figure 1. From the zoomed trace of Figure 6c, *C* reaches V_H faster using the parallel topology. From Equation (10), when $V_C(0) = 0$, the time needed for *C* to reach $V_H = +5$ V is 847.5 s, using the series topology, and 183.1 s, using the parallel topology. When $V_C(0) = +2.23$ V, these times are reduced to 483.4 s and 108.1 s, respectively. In our case, the parallel topology is the best option, because the priority is that *C* reaches V_H faster, not reaching a higher voltage.

The proposed wearable system and its performance are shown in **Figure 7**. The energy-conditioning system, the PMC, and the circuits of Figure 4 were all included in the printed circuit board (PCB) (Figure 7a). Figure 7b shows the performance of the system, where *C* was initially charged to +5.15 V. While VR was off, *C* discharged very slowly due to the low current (≈103 nA) sunk by $R_1 - R_4$ and due to $I_{D\_leakage}$. When VR was turned on, +2 V was supplied to the load, and the current consumption ($= C{\cdot}\Delta V_C/t$) increased to 8 μA until $V_C(t)$ reached V_L (= 2.23 V); the main contribution to this "high" current was due to I_{LOAD} and the current consumption of VR. *C* discharged faster because of this high current; even so, it was enough to maintain the autonomy of the PMC and the load for more than 20 s. During this time, the cardiac pulse signal was clearly obtained. Once $V_C(t)$ reached V_L, VR turned off, and the current consumption decreased again. The theoretical results were obtained from Equation (15), using $V_C(0) = +5.15$ V, $C = 68\,\mu F$, $R_{eq} = 45$ MΩ, $R_{eq\_VD} = 57$ MΩ, and $R_{C\_Leakage} \approx 200$ MΩ.

Figure 7c shows what happened when $V_C(t)$ reached V_L, and then the volunteer wearing the glove closed and opened the hand at 2 Hz; the five sensors were interconnected in parallel. When VR turned off, $V_C(t)$ was at +2.25 V. Assuming this voltage as $V_C(0)$, from Equation (10), the time needed to reach V_H was about 104 s, and the energy stored in *C* was about 850 μJ. The experimental results validate Equations (12) and (15).

Figure 7d shows how *C* was charged from the finger motions when the subject used a computer and a cellular phone for 10 min each. VR was off. When the volunteer used the computer keyboard and mouse, the charging current was about 195 nA, $V_C(t)$ reached +4.2 V, and, according to Equation (12), the stored energy was about 580 μJ. When the subject used the cellular phone, the charging current was 221 nA, $V_C(t)$ reached +4.5 V, and 700 μJ was stored. It is evident that under these conditions, *C* needed more time to reach V_H because the movement of the fingers is not as large as opening and closing the hand at a fixed frequency.

7. Conclusion

A self-powered wearable system that harvests and manages the biomechanical energy of the fingers was proposed. Five PVDF

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film-type sensors mounted on a glove-type garment were used to harvest the energy from finger movement, not the impact on a surface. The harvested energy was conditioned and stored in a 68 μF ceramic capacitor. This energy was used to power for 20 s a battery-less PMC and a battery-less circuit that detects the cardiac pulse of the subject. The PMC was designed to isolate the capacitor from the load to reduce the current consumption while charging. Once the voltage across C reaches a high threshold voltage, the user controlled the energy delivery to the load. From daily activities like using a computer/laptop or a cellular phone for 10 min, it was possible to store up to 700 μJ in the capacitor. Nevertheless, the proposed wearable system is not limited to the activities presented here, because it can harvest the biomechanical energy from any activity in which finger movements are involved.

Supporting Information

Supporting Information is available from the Wiley Online Library or from the author.

Conflict of Interest

The authors declare no conflict of interest.

Data Availability Statement

Research data are not shared.

Keywords

battery less, biomechanical energy, energy harvesting, power management circuits, wearable systems

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