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Energetic Analysis for Self-Powered Cochlear Implants

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Abstract— **Cochlear implants (CIs) are used for compensating the so-called deep sensorineural deafness. CIs are usually powered by rechargeable or long-lasting batteries. In this paper, the feasibility of a fully implanted stand-alone device able to provide the electric power required for stimulating the auditory nerve, without external recharging, is investigated. At first, we demonstrate that the sound wave entering the ear is not a sufficient power source. Then, we propose a solution exploiting the mechanical energy associated to head vibration during walking. The energetic feasibility of this approach is demonstrated based on experimental measurements of head motions. Preliminary considerations on the technical feasibility of a fully implanted energy harvester are finally presented.**

I. INTRODUCTION

Sensorineural deafness is a specific type of severe deafness originating from irreversible damage to the cochlear hair cells [1]. Cochlear implants (CIs) are widely used for the treatment of the severe sensorineural hearing loss [2], [3] in that they bypass the damaged hair cells and directly stimulate the auditory nerve.

A common CI can be divided in two parts [2]. The external part of the CI, positioned behind the pavilion, comprises: a *microphone*, a *speech processor*, a *battery* and a wireless *emitter*. The internal part of the CI, implanted in the inner ear, comprises: a *receiver* and the *electrodes* that stimulate the auditory nerve.

To date, the CIs are powered by batteries [4], [5] that need periodic replacement after exhaustion. Another energy supply approach under investigation looks at the use of inductive powering [4]. The main limit of this approach is the long recharging process that must be repeated daily. None of the above mentioned powering strategies (i.e. batteries or inductive powering) allows for a fully implantable CI, which, in turn, appears to be the optimal solution from the users' point of view, both in terms of comfort and cosmetic. Towards this goal, a possible energy alternative could consist in energy harvesting devices able to convert mechanical energy available in the environment surrounding the CI into electrical energy. In detail, we want to analyze the feasibility of a fully implanted standalone device able to provide the electric power required for stimulating auditory nerve without external recharging.

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II. STATE OF THE ART

A. Energy requirements

In order to estimate the energy requirements for stimulating the auditory nerve using a CI, we have analyzed the Nucleus CI24RE receiver-stimulator, connected to the Nucleus 22-electrode intracochlear array (by *Cochlear*^{TM}, Melbourne, Australia), already considered in a study on electrode impedances [6]. The average impedance between electrodes and tissue is about $4.17 \times 10^3\ \Omega$ [6]. Moreover, the electric power needed to stimulate the auditory nerve during a normal conversation, assuming a linear relation between the intensity of the sound and the current, is $4.17 \times 10^{-3}\ W$.

B. Powering techniques

A common drawback of Active Implantable Medical Devices (IMDs) is their short energetic autonomy. A possible strategy to overcome such a limit consists in recharging an energy storing device, such as a battery or a capacitor, by means of an electromechanical device that converts some form of energy present in the environment into electrical energy. The process by which energy is transduced into the electric domain and stored is conventionally named *Energy Harvesting* or *Energy Scavenging*.

Energy scavengers for IMDs [4] can be classified according to the type of input energy or according to the energy transduction method employed. Several energy sources can be used as input by energy scavengers. In particular, current micro electromechanical systems (MEMS) technology allows to exploit, among the others, vibrations [7] and thermal gradients [8]. A few examples of different energy transduction strategies include: *Thermoelectric*, *Piezoelectric* and *Electromagnetic*. Stark [9] developed the thermoelectric *Thermo Life*. This is a compact energy source, providing about $(10-100) \times 10^{-6}\ W$. Yang et al. [8] investigated the energy generation from an implanted thermoelectric generator (TEG), which exploits the thermal gradient between the inner body and the skin surface. In their in-vivo experiment, the TEG is implanted in a rabbit, the temperature difference is $5.7\ K$ and the output voltage is $25 \times 10^{-3}\ V$. The thermo-power of a single thermocouple can realize a value of about $97 \times 10^{-6}\ V\ K^{-1}$. There are several miniature energy harvesters which use piezoelectric materials to convert mechanical energy, usually available as vibrations, into electricity. To our knowledge, no such device has been implanted into the human body. On the contrary, there are several examples of extra-corporeal use. For instance, Lawrence C. Rome et al. [10] have developed a suspended-load backpack, that converts mechanical energy, related to the vertical movements of carried loads during normal walking, into electrical power. Mukherjee et al. and Franks et al. [5], [11], [12], have studied the possibility to use piezoelectric benders, in the cochlea, to convert the acoustic power in electric power for stimulating the auditory nerve. In this example sound itself is the only energy source. However, we can demonstrate that this approach is not a viable solution. The energy associate to a sound wave is not sufficient to power any CI. Indeed, the relation between pressure (p) and acoustic power (P_{ac}) [13] is:

$$P_{ac} = \frac{p^2 T}{Z} \tag{1}$$

where T is the surface crossed by the wave and Z is the acoustic impedance of the medium. For air, $Z = 444\ kg/m^2\ s$ [13]. Being $T = 85 \times 10^{-6}\ m^2$ [14] the area of the tympanic membrane, the power associated to sound during a normal conversation [13] ($p = 0.02\ Pa$) and available to power a CI is about $76 \times 10^{-12}\ W$. This extremely low power level makes this powering technique practically ineffective to stimulate the auditory nerve: energy must be recovered from other sources. Von Buren et al. [7] carried out a study about the optimization of energy harvesters activated by human walking motion. They compared different motion-powered harvesters using acceleration data of 9 body segments during walking. Their study demonstrated a potential power output up to $1500 \times 10^{-6}\ W$. In this paper, we analyze the feasibility of a fully implanted standalone device able to convert the natural movement of the human head into electric power, required for stimulating auditory nerve without external recharging.

III. ENERGETIC FEASIBILITY ANALYSIS OF AN IMPLANTABLE ENERGY HARVESTER

In this paper we take into account an electromagnetic energy harvester converting into electricity the kinetic energy of a seismic mass, similarly to what is reported in [7], [15]. In our study we theoretically investigated the expected power output when the device is implanted close to the ear and oscillations are due to normal walking.

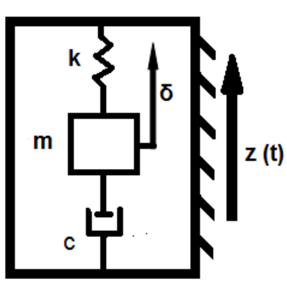

Fig. 1. Generic model of an inertial generator.

The generator is a second-order, mass (m)-spring (k)-damper system (c) (Fig. 1). The input of the system is the motion of the head applied to the frame, $z(t)$, which produces an absolute movement of the seismic mass, $q(t)$. The differential equation of motion for the system is:

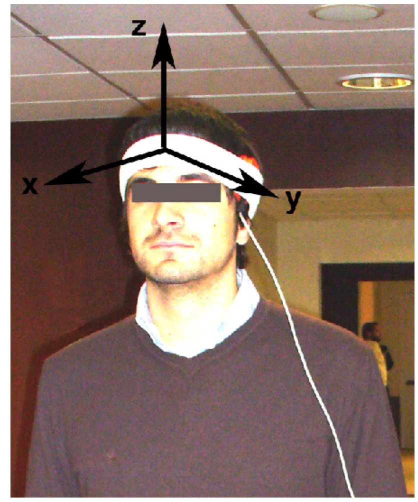

Fig. 2. Reference system for the measurement of the head acceleration. The z-direction corresponds to the vertical direction, pointing upward. The x-direction is horizontal in the walking direction; the y-direction is chosen so to complete a reference frame with positive orientation.

$$m\ddot{\delta} + c\dot{\delta} + k\delta = -m\ddot{z} \tag{2}$$

where δ, $\dot{\delta}$ and $\ddot{\delta}$ are respectively the relative displacement, velocity and acceleration between frame and seismic mass in the time domain. The energy converted into electricity, being irreversibly subtracted from the system, is a fraction α of the total dissipated power by the damper. Evidently, α depends on the specific embodiment of the device, i.e. its detailed design, materials and fabrication technologies. At this stage of the study we assume that α is a constant ($0 < \alpha < 1$) and we choose the dynamic parameters (m, k and c) so to optimize the dissipated power on the damper.

A. Data collection

The head acceleration is measured using a Xsens MTx [16], a 3 DOF orientation tracker, fixed over the ear by an elastic band (Fig. 2).

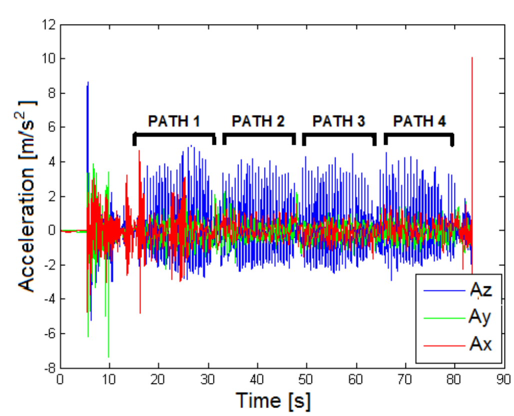

Fig. 3. Example acceleration waveform.

Ten healthy subjects, age: $22 \div 28$ years, were asked to comfortably walk along a $23\ m$ linear path. The walk was repeated four times for each subject, to let them get confident with wearing the sensor. The average stride frequency resulted to be $1.8 \pm 0.4\ Hz$ (this result is in line with [17]). Only the z-component of the three-dimensional acceleration signal (Fig. 3) was taken into account because acceleration is expected to have the highest amplitude along this direction.

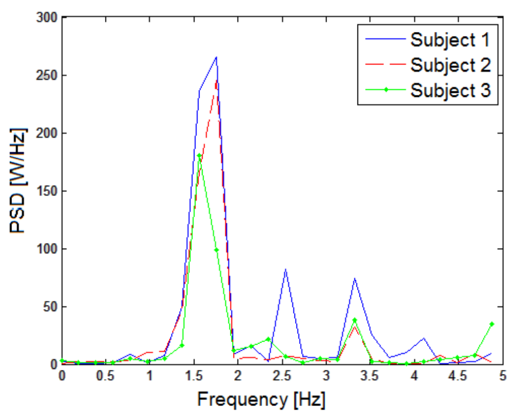

Fig. 4. PSD of the head acceleration along the z-axis which describes how the power of a signal is distributed with frequency.

For each subject, the acceleration in a $15\ s$ time window was analyzed, calculating the power spectrum density (PSD) [18]. PSD describes how the power of the acceleration signal is distributed among frequencies.

The dissipated power on the damper is:

$$p_c = c\dot{\delta}^2 \tag{3}$$

In frequency domain δ, $\dot{\delta}$ and $\ddot{\delta}$ are represented as Δ, $\dot{\Delta}$ and $\ddot{\Delta}$. Analogously, Z and $\ddot{Z}$ respectively represent z and $\ddot{z}$. The instantaneous electrical power, p_e, produced by the system is: $p_e = \alpha\ p_c$. By introducing the transfer function $H(s) = 1/(ms^2 + cs + k)$, we have:

$$\Delta = H(s)\ F(s) \tag{4}$$

where $F(s) = m\,\ddot{Z}(s)$. Obviously, the Laplace transform of $\dot{\delta}$ is: $\dot{\Delta} = s\ H(s)\ F(s)$. Being $s = i\ 2\pi\ f$, the total dissipated power ($\bar{P}_c$) is:

$$\begin{aligned}\bar{P}_c &= 2\int_0^{\infty} P_c(f)\ df = \\ &= 2\int_0^{\infty}\left[c\ m^2\left|i\ 2\pi f\ H(f)\right|^2\left|\ddot{Z}(f)\right|^2\right]df\end{aligned} \tag{5}$$

and the electric power ($\bar{P}_e$) is:

$$\bar{P}_e = \alpha\bar{P}_c \tag{6}$$

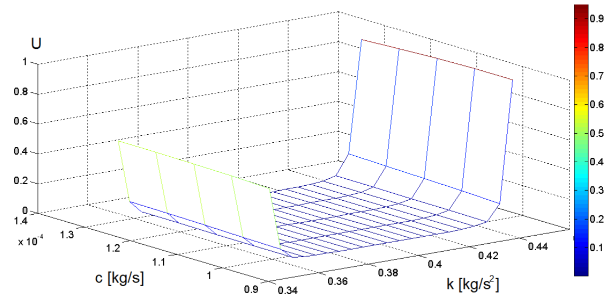

Fig. 5. The function U in terms of c and k. U is maximum for $c = 9.82 \times 10^{-5}\ kg/s$ and $k = 0.45\ kg/s^2$.

B. Optimization

We set the value of the seismic mass to $m = 3.7 \times 10^{-3}\ kg$. Moreover, we chose the the natural frequency of the system to be very close to the measured stride frequency ($\hat{f} = 2\ Hz$). The parameters k and c are provided by the following expressions:

$$\omega_n = \sqrt{\frac{k}{m}} = 2\pi\ \hat{f} \tag{7}$$

$$\xi = \frac{c}{2mw_n} \tag{8}$$

where ω_n is the resonance pulse and ξ the damping factor. In connection with the aforementioned Eq. 5 the available input $PSD = \left|\ddot{Z}(f)\right|^2$ is transmitted through the filter $H(f)$, which depends by c and k. In order to maximize $\bar{P}_c$ for each subject, $H(f)$ should maximize $P_{max_i} = \int_0^{\infty}\left|\ddot{Z}_i(f)\right|^2\ df$ for $i = 1..N$, with $N = 10$. P_{max_i} depends on the stride frequency that varies from subject to subject (see Fig. 4). Since the same device (and so the same $H(f)$) has to be used with all the subjects, a single filter has to be chosen. The values of k and c, which minimize the percentage loss of maximum dissipated power, calculated for each subject, were determined. The parameters that best suit different users are the ones that maximize the following target function:

$$U = \frac{1}{N}\sum_{i=1}^{N}\frac{P_i}{P_{max_i}} \tag{9}$$

where P_i ($i = 1..N$) is a set of (non maximal) power values obtained when the model parameters are constant for the whole set of subjects. By numerical investigation it is easy to show that the quantity U is maximized for $c = 9.82 \times 10^{-5}\ kg/s$ and $k = 0.45\ kg/s^2$ (Fig. 5).

The average dissipated power ($\bar{P}$) is:

$$\bar{P} = \frac{1}{N}\sum_{i=1}^{N} P_{ott_i} \tag{10}$$

where P_{ott_i} is the power calculated fixing $c = 9.82 \times 10^{-5}\ kg/s$ and $k = 0.45\ kg/s^2$.

The electric power ($\overline{P_{elet}}$) is simply given by:

$$\overline{P_{elet}} = \alpha\bar{P} \tag{11}$$

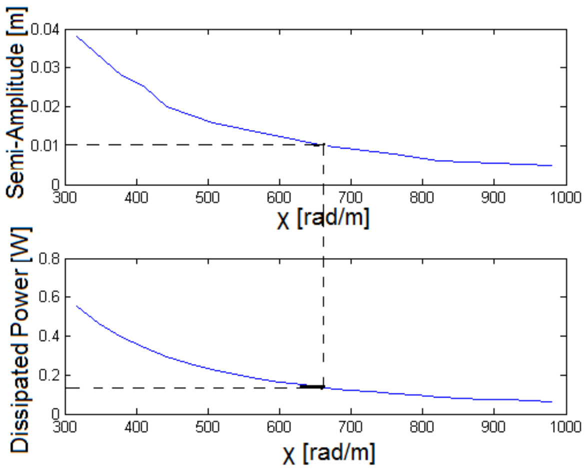

Fig. 6. The dissipated power and the semi-amplitude of the displacement between the mass and the frame in terms of χ.

IV. Results and Discussion

The general monoaxial model, described above, predicts a dissipated power up to $6.44 \pm 2.61\ W$. The actual amount of electric power could be a small fraction α of this value. For instance, by considering $\alpha = 0.03$, a reasonable value given the small dimension of the electromechanical system, the average electric power $\overline{P_{elet}}$ obtained is of $0.193 \pm 0.002\ W$. In order to realize a implantable device, we have supposed a particular configuration of device in which the translating mass is connected to a rotor via a suitable transmission. The translation δ of the mass, due to the head acceleration, causes the rotation of the rotor of an angle $\vartheta = \chi\ \delta$, where χ is the transmission ratio. Assuming the rotor coupled to a torsion spring, the dynamics of the system is described by:

$$\left(m + I\ \delta^2\right)\ \ddot{\delta} + c\ \chi^2\ \dot{\delta} + k\ \chi^2\ \delta = F \tag{12}$$

where $F = -m\ \ddot{z}$ describes the external input, $\ddot{z}$ is the acceleration of the head along z-axis and I is the inertial moment of the rotor. The dissipated power is:

$$p_c = M_c\ \dot{\vartheta} = c\ \dot{\vartheta}\ \dot{\vartheta} = c\ \dot{\vartheta}^2 = c\ \chi^2\ \dot{\delta}^2 \tag{13}$$

By analyzing the system in the frequency domain, one finds that electric power P_e is:

$$P_e(s) = \alpha P_c(s) = \alpha c\ \chi^2 |s\ H(s)|^2\ |F(s)|^2 \tag{14}$$

Fig. 6 shows the dissipated power and the semi-amplitude of the displacement between the mass and the frame as a function of χ. We see that when $\chi = 664.1\ rad/m$ and when the semi-amplitude of displacement is $0.01\ m$, the dissipated power is of $0.14\ W$, so that the electric power is about $4.2 \times 10^{-3}\ W$ $(\alpha = 0.03)$.

V. Conclusions and Future Works

A. Conclusions

The power associated to the sound wave impacting the tympanic membrane is 9 orders of magnitude lower than the electric power necessary to stimulate the auditory nerve using a CI. A different strategy to power such prostheses consists in harvesting energy from the natural movement of the head using an implanted device. The head acceleration data along the z-axis have been measured on 10 test subjects. The dynamic parameters have been optimized by using a second order model. A rotary version of the device, with the seismic mass attached to a rotor, allows the achievement of the necessary small dimensions. In this configuration, the system is likely to provide a sufficient level of electric power, able to stimulate the auditory nerve, when the semi-amplitude of this displacement between the mass and the frame is $0.01\ m$.

B. Future Works

The inertial generator provides intermittent power, that must be conveniently stored. The required capacity is the bigger the shorter are the periods of energy production. In order to keep the overall dimensions compatible with full implantability, a more detailed dimensioning will be performed taking into account the head movements not only during walking but also during other common daily activities.

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