Development of an energy harvesting device with a contactless plucking mechanism driven by a skeletal muscle

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Abstract
We propose an energy harvesting device driven by the contraction of an electrically-stimulated skeletal muscle that can be used as an alternative to batteries for implantable medical devices. In order to realize a durable generator, this device has a contactless plucking mechanism comprising parallel leaf springs and magnets, with which the generator can be driven without friction. By utilizing this mechanism, the generator can be driven not only in the contraction phase of the muscle, but also during relaxation. For the design we optimized the stiffness of the springs, the gap between the magnets, and the magnetic circuit in order to maximize the power generated. The power generated by a prototype in a benchtop experiment was 35.8 μW, which is sufficient to drive an implantable medical device. Furthermore, we evaluated the power generated in an ex-vivo experiment in which the gastrocnemius muscle of a toad weighing 193.4 g was electrically stimulated to drive the mechanism. In this experiment, 18.1 μW of power was generated from a skeletal muscle weighing only 3.5 g. It was also confirmed that the power generated exceeded the power required to electrically stimulate the skeletal muscle. The results showed the feasibility of an energy harvesting system utilizing the proposed mechanism.

Keywords : Energy harvesting, Skeletal muscle, Electrical stimulation, Contactless plucking mechanism, Parallel leaf spring

1. Introduction

The aging populations in the developed countries mean that there are many people suffering from cardiovascular diseases, such as arrhythmias and bradycardia. This has led to a high demand for implantable medical devices (IMDs) such as artificial cardiac pacemakers, implantable cardiac defibrillators (ICD), which are used to accurately control the heart rate, and neurostimulators to reduce chronic back pain and cure dystonia. In general, the power to drive these devices is supplied by batteries, and the power required is 30 – 100 μW for a pacemaker and an ICD and 30 – several mW for a neurostimulator (Schmidt and Skarstad, 2001). Periodically, surgery is required to replace these batteries as they reach the ends of their lifetimes. This surgery is conducted every 10 years for a pacemaker (Mallela et al., 2004), 5 – 9 years for an ICD (Kindermann, et al., 2001; Burleson et al., 2012), and 8 years for a neurostimulator (Medtronic CO., 2017). Despite these long cycle times, replacement surgery is a physical and financial burden for the patient, and there is also a risk of complications (Gould and Krahn, 2006).

It has been reported that the number of interventions, such as surgery, is increasing as the global burden of disease increases (Ozgediz et al., 2008). This surgery often requires long-term care in hospital to monitor postoperative complications such as surgical site infection, inadequate tissue healing, and sepsis (Venema et al., 2014). This increases the health care cost, which is a big financial burden for the patient. It has also been reported that the number of elderly suffering from chronic diseases such as diabetes is increasing (Jagger et al., 2007). Life-threatening events caused by chronic diseases generally occur infrequently, and traditional monitoring is limited to brief time points, which may not be sufficient to detect the disease, resulting in delayed diagnoses (Perez et al., 2015). Therefore, continuous health monitoring is in high demand. Thus, interest in the development of wearable or implantable health monitoring devices
utilizing sensors that provide continuous measurement of the physiological state of a person is increasing. Accurate sensors for blood pressure, pulse rate, cardiac rhythm (Tsouri and Ostertag, 2014), and blood glucose (Lucisano et al., 2017) have already been developed. Especially, arrhythmias can be detected in near real time and a patient can monitor his own physiological state on a smartphone (Adibi, 2014). However, one problem with these devices is maintaining the energy resource. According to work done by Nia et al. in 2015, the power consumed in measuring the heart rate and sending the data is 17.36 – 47.11 μW, and that for measuring blood pressure requires 7.52 – 803.01 μW. Assuming each sensor is powered by the regular coin cells commonly used in biomedical sensors, the battery would run out in a few days. Therefore, an energy resource that can supply long term power is essential not only for IMDs but also for continuous health monitoring.

One of the solutions to these problems is to supply electrical energy generated from some resource in the human body. This requires an implantable energy harvesting system. A study related to implantable energy harvesting systems was reported in 1963 by Parsonnet. He developed a device which converts the kinetic energy produced by the pulsating aorta into electrical energy using the piezoelectric effect. Nowadays, many types of implantable energy harvesting device using different resources and conversion methods are being researched.

The glucose biofuel cell (GBFC) is a device in which an enzymatic reaction converts the chemical energy of glucose in human body fluids to electrical energy. This conversion method is considered to be the most efficient method since the flow of energy in the human body is mostly chemical energy. The GBFC developed by Zebda et al. in 2013 achieved a delivered power of 38.7 μW. Despite the high output, the lifetime of a GBFC is quite short due to the enzymes, whose lifetime is at most about a month (Moore et al., 2004). Also, implantable photovoltaic energy harvesting devices that generate power from sunlight or artificial light have been proposed. The device developed by Wu et al. in 2018 achieved 2.75 mW momentarily when exposed to direct sunlight. However, the power generated by this type of energy harvesting system depends on external factors such as the weather and the time of day. In addition, it is necessary that the device be exposed to direct sunlight (or light whose brightness is equivalent to it) for 1 – 2 hours in order to power an IMD. For these reasons, the patient's activity is possibly limited. Other resources in the human body are thermal energy and kinetic energy. In 2013, Leonov developed a thermoelectric generator utilizing the temperature difference between the body and the ambient environment. A prototype achieved a power density of 571 μW/cm³, however, in the case of implantation of this device, the power generated may be much less due to the small temperature differences within the human body. Energy harvesting devices utilizing the kinetic energy of a beating heart (Tashiro et al., 2002; Zurbuchen et al., 2013), arterial deformation (Pfenniger et al., 2013; Sohn et al., 2005), and blood flow (Wang et al., 2012) have been developed. However, since the heart and arteries (or blood vessels) are essential to maintaining life, there are risks involved in implanting these devices close to the heart and the arteries, and the possible breakdown of these devices may cause, possibly fatal, damage.

Another form of kinetic energy in the human body is the contraction of a skeletal muscle. In our work, we have focused on an energy harvesting system utilizing electrically stimulated muscle contraction as shown in Fig. 1. Skeletal muscles are voluntary muscles; however, they can be forced to contract by electrical stimulation supplied from the battery of an IMD, and then used to continuously generate energy. The energy harvesting device converts the kinetic energy of the contraction into electrical energy. A schematic of the proposed energy harvesting system is shown in Fig. 1. The principle of operation is as follows: electrical stimulation is delivered to the muscle via stimulation electrodes, driving the muscle to contract. The contraction generates mechanical energy, which is then converted to electrical energy by a generator. The produced electrical energy can be stored in an implantable battery and used to power an IMD.
energy of the contraction into electrical energy, which is then supplied to the battery. Most of the electrical energy is consumed by the IMD, the remaining energy is used to continue stimulation of the skeletal muscle. The advantages of this system are; the surgery required to implant the generator on a skeletal muscle is easier than on the heart or an artery; in the case of failure of the device, the risk to the individual's life is small. Also, the power density generated by contraction of a skeletal muscle is reported to be 1 mW/g (Araki et al., 1995; Gustafson et al., 2006), while the energy consumed for stimulation is some micro watts. Therefore, it is possible that only a small skeletal muscle is needed to supply an implantable medical device such as a pacemaker.

In 2007 and 2009, Lewandowski et al. developed an energy harvesting device using an electrically stimulated muscle and a piezoelectric generator. The device could generate 0.83 μW from 64 g of a rabbit’s quadriceps; however, the power required to stimulate the skeletal muscle meant that the net power generated was only 0.03 μW. The reason for this small net power is due to the difference between the properties of the skeletal muscle and the piezo element. In general, a skeletal muscle contracts by about 1 mm at a few Hz, while the power generated by a piezo element is largely done by displacement by a few micrometers. On the other hand, with electromagnetic induction power is generated by large displacement, so this conversion method is suitable for the mechanical properties of skeletal muscle.

Our research group previously developed a prototype electromagnetic generator utilizing the contraction of a skeletal muscle (Sahara et al., 2016). With this a power of 134 μW was generated from 1.3 g of the skeletal muscle of a toad, and the net power generated was 111 μW. Although this showed the feasibility of the proposed system, this prototype had a problem in terms of durability. This was because some parts, such as the gears and bearings, required periodic maintenance, so long-term usage of this prototype in the human body was not feasible.

The purpose of this paper is to realize an energy harvesting system which utilizes the contraction of a skeletal muscle that can be used permanently in the human body. Thus, we propose using a contactless generator to improve durability. The proposed generator is designed to maximize the power generated. Based on the design, a prototype was fabricated and the power generated in a benchtop experiment was evaluated. Then, the prototype was demonstrated in an ex-vivo experiment utilizing the gastrocnemius muscle of a toad.

2. Proposed contactless energy harvesting device

The proposed contactless energy harvesting device is shown in Fig. 2. There are two parts to the device, a plucking mechanism and an electromagnetic induction part or generator. The plucking mechanism consists of permanent magnets, magnetized in opposite directions to each other, and two parallel leaf springs. The parallel leaf springs and the magnets for the plucking mechanism are divided into a drive part which is connected directly to a skeletal muscle and an oscillator which is driven by the repulsive force between the magnets in the mechanism. The oscillator consists of the driven magnet, two parallel leaf springs and coils. The generator converts the kinetic energy of the oscillator into electrical energy by electromagnetic induction. In general, the power generated is proportional to the second power of the natural frequency. Thus, in the electromagnetic induction part, the coil, which weighs less than the permanent magnets, is attached to the driven parallel leaf springs and the permanent magnets are mounted on the stator. Additionally, the coil is located both in front of and behind the permanent magnets in order to improve the power generation efficiency. Figure 3 illustrates a particular arrangement of the energy harvesting system with a pacemaker, which we proposed previously (Sahara et al., 2016). This system consists of a generator, electrodes, an electrical circuit, a storage battery, a shell, and a wire to transfer the power of the muscle to the generator located on the greater pectoral muscle. The electrical circuit includes a signal generator for stimulation, a rectifier for the generated power, and a circuit to monitor the battery level of the pacemaker. To operate the system, muscle fiber in the greater pectoral muscle, which is long enough to drive the generator, is cut, and the end of it is connected to the generator through a wire contained within a tube.

Usually the generator is driven by movement of the patient. However, when the monitoring circuit detects that the battery level is low, electrical stimulation is applied to the muscle, and the contraction of the muscle is used to drive the generator.
Figure 4 illustrates the driving principle of the energy harvesting device. The skeletal muscle contracts due to electrical stimulation and pulls the drive springs as shown in Fig. 4(a). $F_{\text{muscle}}$ is the contraction force of the skeletal muscle and $F_{\text{elastic}}$ is the elastic force in each parallel leaf spring. The repulsive force $F_{\text{mag}}$ between the magnets in the plucking mechanism drives the other pair of springs. When the elastic force of the driven parallel leaf springs exceeds the repulsive force between the magnets, the oscillator starts to vibrate freely. This mechanical energy is converted into electrical energy in the electromagnetic induction part. Since this is a contactless mechanism, there is no friction or abrasion between the moving parts, and the generator is expected to be highly durable.

After the energy has been converted and the stimulation has been stopped, the muscle relaxes, and the parallel leaf springs move back to their initial positions due to the elastic forces in each of them, as shown in Fig. 4(b). At this moment, when the plucking mechanism has been released, the oscillator starts to vibrate again. The advantage of this mechanism is that energy conversion occurs twice, i.e., in the contraction phase and when the muscle relaxes, with a single stimulation of the skeletal muscle. Since, with this plucking mechanism the conversion efficiency can be increased, the stimulation cycle time can be increased, which is expected to reduce fatigue in the muscle.

3. Design of the energy harvesting device

3.1 Design goal

The design requires appropriate values for the stiffness of each spring, and the gap and initial offset between the magnets in the plucking mechanism. If these parameters are inappropriate, the plucking mechanism will not function and the oscillator will not vibrate when the muscle contracts and relaxes. In addition to these parameters, the design of the magnetic circuit in the electromagnetic induction part is important to maximize the power generated. In this section, we describe the design of the magnetic circuit, and the procedure used to choose the values for the stiffness, and the gap and initial offset between the magnets in the plucking mechanism.

![Fig. 4 Principle of proposed plucking mechanism](image-url)
The configuration of the generator is shown in Fig. 5. Considering that this is to be implanted in a human body, the size of the generator needs to be as small as possible. Thus, the size of the generator was reduced by arranging the two sets of parallel leaf springs to be the same size, rather than having the drive springs on either side of the driven springs. The target value for the average power is 30 μW, which is the least power required to drive an IMD, and the target volume is the same as that of the battery, which is about two thirds the size of the pacemaker. For example, the outer dimensions of the pacemaker (ACCOLADE MRI™, Boston Scientific Japan Co., Tokyo, Japan) are 44.5 mm × 58.8 mm × 7.5 mm, which is 1.58×10⁻⁵ m³. Hence, the target volume is about 10⁻⁵ m³.

3.2 Design of the magnetic circuit

The coil has a meandering shape with a pitch of 1.0 mm, and is made of six-stranded 0.12 mm diameter copper wire, as shown in Fig. 6. The magnets for the generator are 1 mm × 1 mm × 20 mm neodymium-iron-boron magnets (Magfine Co, Miyagi, Japan). Two types of magnetic circuit in the electromagnetic induction part were considered; in...
type 1, the magnets are fixed in the grooves of an aluminum core as shown in Fig. 7(a); in type 2, the magnets are fixed to an iron core such that the magnetization in adjacent magnets is in opposite directions, as shown in Fig. 7(b). In type 2, the magnetic flux density generated from the magnets will be larger than that in type 1. However, the magnetic flux across the coil in type 2 will be smaller than that in type 1 since the magnetic circuit between adjacent magnets is closed. Hence, we compared the magnetic flux density $B$ across the coil in the $z$ direction for each type using magnetic field analysis software (JMAG, JSOL Co., Tokyo, Japan). The magnetic flux density across the coil was simulated with the coil displaced from 0 mm to 2 mm in the $y$ direction. The values of the gap between the magnets and the coil were varied from 0.1 to 0.3 mm, chosen by considering the accuracy with which the device could be manufactured and assembled.

The simulated magnetic flux densities across the coil in the $z$ direction are shown in Fig. 8(a) and (b). In type 1, even when the gap is 0.1 mm, the maximum value of the magnetic flux density is only 0.17 T. On the other hand, this is more than 0.2 T with a gap of 0.3 mm and 0.28 T with a gap of 0.1 mm in type 2. Based on these results, we adopted the type 2 magnetic circuit for the energy harvesting device.

### 3.3 Optimization of the stiffness

In order to maximize the power generated by the type 2 magnetic circuit with an air gap of 0.1 mm, we optimized the stiffness, the gap between the magnets in the plucking mechanism, and the initial offset between the magnets. The generated power can be calculated by solving the equations of motion for the parallel leaf springs, which are as follows:

$$m_1 \ddot{x}_1 + c_1 \dot{x}_1 + k_1 x_1 = F_{\text{mus}} - F_{\text{mag}}$$

(1)

$$m_2 \ddot{x}_2 + (c_2 + c_{\text{gen}}) \dot{x}_2 + k_2 x_2 = F_{\text{mag}}$$

(2)

where, $m$, $c$, $k$, $x$, are the mass, damping coefficient, stiffness, and displacement of the parallel leaf springs. The subscripts 1 and 2 represent the driving and driven springs, respectively, and $c_{\text{gen}}$ is the damping coefficient due to electromagnetic induction. $F_{\text{mus}}$ is the force developed by contraction of the skeletal muscle and $F_{\text{mag}}$ is the repulsive force acting between the magnets of the plucking mechanism, the value of which is a function of the relative displacement between the driving and driven springs. By measuring the displacement and stiffness of the springs and calculating the relative displacement and the force, $F_{\text{mag}}$ can be derived as follows.

$$F_{\text{mag}} = a(x_2 - x_1 + \delta) \exp(b(x_2 - x_1 + \delta)^2)$$

(3)

where $\delta$ is the initial offset between the drive and driven magnets in the plucking mechanism, and $a$ and $b$ are constants given by

![Fig. 8 Simulated magnetic flux density across the coil in the $z$ direction](image-url)
\[ a = -2.17 \times 10^5 d + 6.50 \times 10^2 \quad \text{(N/m)} \]  
\[ b = -1.55 \times 10^7 d + 5.03 \times 10^5 \quad \text{(m^2)} \]

where, \( d \) is the gap between the drive and driven magnets of the plucking mechanism.

To derive \( c_{\text{gen}} \), the Lorentz force is considered. The voltage \( V \) induced in the coil by the Lorentz force at position \( x_2 \) and velocity \( \dot{x}_2 \) is given by

\[ V = n l B(x_2) \dot{x}_2 \]

where \( l \) is the length of the coil and \( n \) is the number of turns in it as shown in Fig. 6. The power \( P_{\text{out}} \) generated is defined by the energy consumed by a load resistance connected to the coil, and is given by

\[ P_{\text{out}} = \frac{2R_l V^2}{(R_l + R_c)^2} = \frac{2R_l (n l B(x_2))^2}{(R_l + R_c)^2} \dot{x}_2^2 \]

where, \( R_l \) is the load resistance, \( R_c \) the parasitic resistance of the coil, and \( B(x_2) \) the magnetic flux density with a gap of 0.1 mm derived in the previous section. Differentiating \( P_{\text{out}} \) with respect to \( R_l \) gives:

\[ \frac{dP_{\text{out}}}{dR_l} = 2 (n l B(x_2))^2 \dot{x}_2^2 \left( \frac{R_c - R_l}{(R_l + R_c)^3} \right) \]

Thus, the maximum value of \( P_{\text{out}} \), is obtained when \( R_l \) is set to same value as \( R_c \). In this case, \( P_{\text{out}} \) is given by

\[ P_{\text{out}} = \frac{(n l B(x_2))^2}{2R} \dot{x}_2^2 \]

Fig. 9 Results of calculations for design optimization
Since the power generated is a product of the damping force $F_d (= c_{gen} \dot{x}_2)$ induced by electromagnetic induction and the velocity $\dot{x}_2$, $c_{gen}$ is derived as follows.

$$c_{gen} = \frac{F_d}{\dot{x}_2^2} = \frac{P_{out}}{\dot{x}_2^2} = \left(\frac{nlB(\dot{x}_2)}{2R}\right)^2$$

(10)

In this simulation, the force due to contraction of the skeletal muscle $F_{mus}$ is the input. This is a function of $t$ such that its value increases to 1.5 N with a gradient of 13.1 N/s, is held at 1.5 N for 5 s, and then decreases with a gradient of 2.73 N/s. The values of the force gradients were determined by experimental measurements of force on the skeletal muscle of a toad, whose length and weight were 30 mm and 3.8 g, respectively. From Eqs. (1) - (5) and (10), the response of each parallel leaf spring and the power generated were simulated using numerical calculation software (Simulink, The Math Works Inc., Natick, MA, USA). By changing the design parameters $k_1$, $k_2$, $d$, and $\delta$, we explored the optimal parameters to maximize the power generated. For these calculations, we assumed electrical stimulation to contract and relax the muscle was supplied for 1.0 s in each case. Thus, the average power required to stimulate the muscle was calculated assuming power was supplied for 2.0 s.

The results of the calculations are shown in Fig. 9. In the black region, the plucking mechanism is not released because one or other of the springs is too stiff. Since the stiffness is proportional to the cube of the thickness of the spring, and the accuracy with which the thickness can be controlled may not be very good, it is best if the power generated is not too sensitive to stiffness. The sensitivity is related to $d$ and $\delta$. When changing $d$ (Figs. 9 (a), (b), and (c)), the sensitivity to stiffness of the power generated is smallest when $d = 1.0$ mm. When changing $\delta$ (Fig. 9 (b), (d), and (e)), the sensitivity to stiffness of the power generated is smallest when $\delta = 1.2$ mm. Thus, we chose these values for $d$ and $\delta$, i.e. the results given in Fig. 9 (b). In this figure, the highest power generated is 211 μW, which is with $k_1 = 600$ N/m and $k_2 = 205$ N/m. Therefore, we set the stiffnesses of the springs to these values.

3.4 Determining the size of the parallel leaf springs

The dimensions of the parallel leaf springs were designed using the optimized stiffnesses derived in the previous section. The relationship between the dimensions of a spring and its stiffness is expressed as follows (Kyusojin et al., 1987).

$$k = Ew\left(\frac{t}{L}\right)^3$$

(11)

where, $E$ is Young’s modulus, and $w$, $t$, and $L$ are the width, thickness, and length of the spring, respectively. Considering long-term usage, the fatigue strength is one of the most important points in the design. Therefore, we used structural analysis software (SOLIDWORKS 2016 Simulation, Dassault Systèmes SolidWorks Corp., Concord, MA, USA).
USA) to design the dimensions of the springs so that the stress acting on them would be less than the fatigue limit. According to the simulation in the previous section, which is shown in Fig. 10, the maximum displacement of the drive spring, $x_1$, is 3.8 mm, and the amplitude of vibration of the driven spring, $x_2$, is 2.6 mm. Applying these displacements, we evaluated the fatigue strength by calculating the maximum stress. We chose beryllium copper C1720-1/4HT (NGK INSULATORS, LTD, Aichi, Japan) as the material for the parallel leaf springs, since it has good stress fatigue performance. According to the fatigue limit diagram of this material, it is known that the fatigue limit becomes flat after $10^8$ cycles of load stress, and the stress after $10^8$ cycles is 880 MPa for pulsative loading, and 460 MPa for alternate loading. The value of the pulsating test was applied to evaluate fatigue in the drive spring since this was displaced from -1.0 to 3.8 mm by contraction. On the other hand, the value of the alternating test was applied to evaluate fatigue in the driven spring since this vibrated in the range ±2.6 mm. If the maximum stress is less than these values, it is expected that the energy harvesting device can be in permanent use.

The results of the calculations of the stress analysis are shown in Fig. 11. When the length, thickness, and width of the drive parallel leaf springs were set to 23 mm, 0.22 mm, and 2.7 mm, respectively, the maximum stress was 549 MPa, which is less than the value obtained from the pulsative loading test, and when the length, thickness, and width of the driven parallel leaf spring were set to 23 mm, 0.17 mm, and 2.0 mm, respectively, the maximum stress acting on the spring was 328 MPa, which is less than the value obtained from the alternate loading test.

Modal frequency of the parallel leaf spring was calculated. The result shows that the first order modal (vibration in the $y$ direction) frequency is 65 Hz, and second order modal (vibration in the $z$ direction) frequency is 333 Hz. In addition, stiffness of the driven parallel leaf spring in the $z$ direction is $7.69\times10^3$ N/m, the value of which is sufficiently larger than that in the $y$ direction. Therefore, it is expected that the vibration in the $z$ direction is sufficiently small when plucking mechanism is released.

Fig. 12 Prototype of generator

Fig. 13 Measured stiffness of the parallel leaf springs
4. Prototyping

Based on the dimensions determined for the design, a prototype was manufactured as shown in Fig. 12. The outer dimensions were 29 mm × 48 mm × 7.7 mm and its volume $1.07 \times 10^{-5} \text{ m}^3$, which is approximately two thirds of the volume of a pacemaker. In order to measure the difference between the designed stiffness values and those of the prototype, we calculated the stiffnesses by measuring the displacements of the parallel leaf springs against an applied load. The calculated stiffnesses, shown in Fig. 13, are 592 N/m for the drive spring and 205 N/m for the driven spring. This confirmed that the stiffnesses of the springs in the prototype are in good agreement with the values in the design.

5. Benchtop experiment to evaluate the power generated

5.1 Experimental method

The experimental set up is shown in Fig. 14. The plucking mechanism was connected via a wire to a stepper motor (HSTM42-1.8, 200 steps/revolution, Changzhou Fulling Motor Co., Ltd, Changzhou, China). The motor simulated contraction and relaxation of the skeletal muscle. We controlled the force generated from the stepper motor by measuring the displacement of the drive springs with laser displacement sensors (LK-G85 (on the left), LK-G155 (on the right), KEYENCE Co., Osaka, Japan) such that the maximum force acting on the springs was 1.5 N. The coils were connected to load resistances with the same values as those of the coils: $R_{L1} = 0.30 \Omega$ for the top coil, $R_{L2} = 1.02 \Omega$ for the bottom coil. We measured the voltages induced by the Lorentz force in the top coil, $V_{L1}$, and the bottom coil, $V_{L2}$, with a voltage follower (LM 741CN, Texas Instruments Inc., TX, USA) and calculated the power generated, $P_{out}$, for periods of 1 s in both the contraction and relaxation phases as follows,

$$P_{out} = \frac{1}{2} \left( \frac{\int_{t_c}^{t_c+1} V_{L1}^2 \, dt + \int_{t_r}^{t_r+1} V_{L1}^2 \, dt}{R_{L1}} + \frac{\int_{t_c}^{t_c+1} V_{L2}^2 \, dt + \int_{t_r}^{t_r+1} V_{L2}^2 \, dt}{R_{L2}} \right)$$

(12)

where, $t_c$ and $t_r$ are the times from which the measurements in the contraction and relaxation phases were taken. In this experiment, the inductance and the capacitance of the circuit were not taken into consideration.

5.2 Result and discussion

The measured displacement of the springs is shown in Fig. 15. The power generated on the top and bottom sides are shown in Fig. 16(a) and (b), respectively. The maximum displacement of the drive springs was 2.93 mm and the calculated force was 1.73 N. This value is slightly bigger than that obtained in the simulation, however, this difference is considered to be acceptable, since fine control of the displacement by the stepper motor was limited by the number of steps. The power generated was 20.3 μW on the top side and 15.5 μW on the bottom side, i.e., the total power generated...
was 35.8 μW.

One reason why the power generated by the prototype was smaller than that predicted by simulation is considered to be the decrease of the magnetic flux density due to the processing of the magnets in the electromagnetic induction part. Another reason is that the damping ratio of the oscillator was much larger than that given by simulation. This was $4.93 \times 10^{-3}$ for the prototype, whereas it was $1.4 \times 10^{-3}$ from simulation.

In order to examine the main cause of the increase in damping ratio, we measured the displacement of the oscillator in three cases; first, free vibration without attachment to the plucking mechanism or connection to the electrical circuit through the coil leads (Case 1); second, with the plucking mechanism, but without connection to the circuit (Case 2); and third, with the plucking mechanism and connection to the circuit via the coil leads (Case 3). From the measured displacements, the damping ratio in each case was calculated. The results are shown in Table 1. The damping ratio in Case 3 is the largest value of the three cases, and this means that the vibration of the wire leads accompanying the vibration of the oscillator was the main cause of the large damping coefficient. In the simulation, the effect of the leads on the damping coefficient was not considered, only the effect of plucking was considered. By arranging the coil on the stator and the magnets for generation on the oscillator (moving magnet type), the power generated may be improved since the damping caused by the leads would be removed. In this case, however, the natural frequency of the oscillator would decrease because of the weight of the magnets. Therefore, we need, in future work, to compare the power generated by a moving coil type with a moving magnet type.

6. Evaluation of power generation in an ex-vivo experiment

6.1 Experimental method

The power generated by the prototype was evaluated in an ex-vivo experiment in order to confirm whether an actual skeletal muscle would be able to drive the generator.

The experimental set up is shown in Fig. 17. The toad was anesthetized with 0.2 wt% tricaine and 0.4 wt% sodium hydrogen carbonate to maintain a neutral pH when immersed. After 30 minutes, confirming that the toad did not react to any external stimulus, we operated on the skin and to reveal the muscle. The legs of the toad were sufficiently well-developed that we expected the muscles would be able to drive the plucking mechanism. Therefore, we used the gastrocnemius muscle of the toad for the skeletal muscle in our experiments. We made an incision in its right leg, cut the Achilles tendon, and removed the muscle from the bone. The gastrocnemius muscle of the toad weighed 3.5 g and its length was 32 mm. After the operation, an S-shape hook and electrodes were threaded into the muscle. Then, the S-shape hook was connected to the generator via a wire. A rectangular wave with a frequency of 30 Hz was applied to

| Case 1 (free vibration of the driven parallel leaf spring) | Damping ratio |
|----------------------------------------------------------|---------------|
| Case 2 (Case 1 with plucking mechanism)                  | $5.98 \times 10^{-4}$ |
| Case 3 (Case 2 with connecting the electric circuit)     | $1.61 \times 10^{-3}$ |
|                                                          | $4.30 \times 10^{-3}$ |
the electrodes for stimulation. In order to minimize the power consumption for stimulation, we measured the impedance of the muscle in advance with an impedance analyzer (IM3570, HIOKI E.E. Co., Nagano, Japan). Based on the measured impedance, the voltage amplitude and width of the stimulation signal were chosen to be 2.49 V and 0.183 ms, so that the predicted power consumption would be about 10 μW. Similar to the benchtop experiment in section 5, the displacement of the parallel leaf springs and the induced voltage were measured, and the power generated was calculated. In addition, since the contraction force of the skeletal muscle was less than 1.5 N, the drive springs were displaced in the positive direction in advance and given pre-tension so that the plucking mechanism could be released with this force.

6.2 Result and discussion

The power consumed by stimulation and the power generated are shown in Figs. 18 and 19, respectively. In the experiment, the muscle was stimulated for 0.5 s in order for the net energy generated, shown in Fig. 20, to become positive. The power generated in 1.0 s was evaluated since the net energy became saturated 1.0 s after stimulation of the muscle. As a result, the power generated was 18.11 μW, while the power consumption due to stimulation was 16.92 μW. Therefore, the net power generated by the prototype in the experiment was 1.19 μW. This confirmed that the generator could be driven by just a small muscle, and that this small muscle could generate electrical energy exceeding the energy needed to stimulate it. The reason why the power generated was smaller than that in the benchtop experiment was the lack of a contraction force and the pre-tension in the wire. The measured displacement of the drive spring shown in Fig. 21 shows that there are two regions: region (i) where the drive spring is driven by the contraction force of the muscle and region (ii) where it is driven by the repulsive force of the magnets since the plucking mechanism was released. The displacement driven by the contraction force of the muscle is about 1.5 mm in region (i).
From this and the stiffness the contraction force was calculated to be 0.9 N, while the force in the benchtop experiment was 1.73 N. Also, pre-tensioning of the wire was needed in the experiment since the contraction force was too small to release the plucking mechanism. In the contraction phase (0 ~ 0.57 s), the gap $g_c$ between the driving and driven springs was so large that the repulsive force between the magnets was negligible and the damping ratio was small. However, the gap $g_r$ in the relaxation phase (0.57 ~ 1.00 s) was smaller than $g_c$, so the repulsive force was not negligible causing kinetic energy to be exchanged between the two springs. During relaxation, since the kinetic energy of the drive spring dissipates through the wire due to the pre-tension, the damping ratio is much bigger than that in the contraction phase.

It was confirmed in this experiment that the net energy generated by the prototype with just a small skeletal muscle was positive, so larger amounts of energy are expected to be generated by the much larger skeletal muscles in the human body. However, some issues to be considered are left for the implantation. Since the wire was connected to a S-shape hook, which was fixed to the skeletal muscle by making a hole on the muscle, there is a possibility that the hole becomes large due to a repetitive load. This may lead the slack of the wire. Thus, change of the muscle around the connection part will be investigated for long-term usage of the generator. In addition, the sealing mechanism for the wire to prevent liquid from entering the generator is required. Since we assume the use of the diaphragm type elastic materials for the sealing, it is required to design the stiffness of the drive parallel leaf spring considering that of the sealing mechanism. Furthermore, the fatigue of sealing mechanism occurred by the repetitive load should also be considered.

7. Conclusion

An energy harvesting system utilizing the contraction of a skeletal muscle, which can be implanted and remain permanently in the human body was proposed. The system consists of an electromagnetic generator and a contactless plucking mechanism comprising parallel leaf springs and permanent magnets. A prototype was designed, and to maximize the power generated, the stiffness of the springs and the gap between the magnets of the plucking mechanism, and the magnetic circuit of the generator were optimized. The power generated by the prototype in a benchtop experiment was 35.8 μW. Moreover, in an ex-vivo experiment the power generated was 18.1 μW and the net power, considering the power required to stimulate the skeletal muscle, was 1.19 μW. Hence, we confirmed that the proposed generator could generate more energy than that needed to stimulate the skeletal muscle. In future work, we plan to increase the power generated in order to drive an IMD and to decrease the energy required to stimulate the skeletal muscle. Since larger skeletal muscles develop stronger forces, the power generated by the larger skeletal muscles in the human body will be greater than that measured in these experiments. Disturbance vibration generated by the motion of the patient’s body may affect vibration of the driven parallel leaf spring. Thus, influence of the disturbance on the driven parallel leaf spring will be evaluated by the vibration test. In addition, fatigue of the muscle...
and change of the skeletal muscle around the connection part will be considered in the design process for long-term usage of the generator.

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References

Abidi, S., Biomedical Sensing Analyzer (BSA) for Mobile-Health (mHealth)-LTE, IEEE Journal of Biomedical and Health Informatics, Vol.18, No.1 (2014), pp.345–351.
Araki, K., Nakatani, T., Toda, K., Tatenaka, Y., Tatsumi, T., Masuzawa, T., Baba, Y., Yagura, A., Wakisaka, Y., Eya, K., Takano, H., and Koga, Y., Power of the fatigue resistant in situ latissimus dorsi muscle, American Society for Artificial Internal Organs Journal, Vol.41, No.3 (1995), M768–771.
Burleson, W., Clark, S. S., Ransford, B., and Fu, K., Design challenges for secure implantable medical devices, Proceedings of the 49th Annual Design Automation Conference (DAC’12) (2012), pp.12–17.
Gould, A. P., and Krahn, D. A., Complications associated with implantable cardioverter-defibrillator replacement in response to device advisories, Journal of the American Medical Association, Vol.295, No.16 (2006), pp.1907–1911.
Gustafson, J. K., Mariache, M. S., Egrie, D. G., and Reichenbach, H. S., Models of metabolic utilization predict limiting conditions for sustained power from conditioned skeletal muscle, Annals of Biomedical Engineering, Vol.34, No.5 (2006), pp.790–798.
Jagger, C., Matthews, J. R., Matthews, E. F., Spiers, A. N., Nickson, J., Paykel, S. E., Huppert, A. F., Brayne, C., and the Medical Research Council Cognitive Function and Aging Study (MRC-CFAS), Cohort differences in diseases and disability in the young-old: findings from the MRC Cognitive Function and Aging Study (MRC-CFAS), BMC Public Health, Vol.7, No.156 (2007), pp.1–8.
Kindermann, M., Schwaab, B., Berg, M., and Fröhlig, G., Longevity of dual chamber pacemakers: devices and patient related determinants, Journal of Clinical Electrophysiology, Vol.24, No.5 (2001), pp.810–815.
Kyusojin, A., Sagawa, D., and Toyama, A., Linear and rotary movement mechanism utilizing leaf springs, Journal of the Japan Society for Precision Engineering, Vol.53, No.7 (1987), pp.1092–1096 (in Japanese).
Lenov, V., Thermoelectric energy harvesting of human body heat for wearable sensors, IEEE Sensors Journal, Vol.13, No.6 (2013), pp.2284–2291.
Lewandowski, E. B., Kilgore, L. K., and Gustafson, J. K., Design considerations for an implantable, muscle powered piezoelectric system for generating electrical power, Annals of Biomedical Engineering, Vol.35, No.4 (2007), pp.631–641.
Lewandowski, E. B., Kilgore, L. K., and Gustafson, J. K., In vivo demonstration of a self-sustaining, implantable, stimulated-muscle-powered piezoelectric generator prototype, Annals of Biomedical Engineering, Vol.37, No.11 (2009), pp.2390–2401.
Lucisano, Y. J., Routh, L. T., Lin, T. J., and Gough, A. D., Glucose monitoring in individual with diabetes using a long-term implanted sensor/telemetry system and model, IEEE Transactions on Biomedical Engineering, Vol.64, No.9 (2017), pp.1982–1993.
Malleta, S. V., Ilhankumaran, V., and Rao, S. N., Trends in cardiac pacemaker batteries, Indian Pacing and Electrophysiology Journal, Vol.4, No.4 (2004), pp.201–212.
Medtronic CO., Restore sensor SureScan MRI™, Package Inserts of Medical Devices (2017), Approval No. 22508BZX00344000 (in Japanese).
Moore, M. C., Akers, L. N., Hill, D. A., Johnson, C. Z., and Minteer, D. S., Improving the environment for immobilized dehydrogenase enzymes by modifying nafion with tetraalkylammonium bromides, Biomacromolecules, Vol.5, No.4 (2004), pp.1241–1247.
Nia, M. A., Kermani, M. M., Kolay, S. S., Raghunathan, A., and Jha, K. N., Energy-efficient long-term continuous personal health monitoring, IEEE Transactions on Multiscale Computing Systems, Vol.1, No.2 (2015), pp.85–98.
Ozgediz, D., Jamison, D., Cherian, M., and Mcqueen, K., The burden of surgical conditions and access to surgical care in low- and middle-income countries, Bulletin of the World Health Organization, Vol.86, No.8 (2008), pp.646–647.

Parsonnet, V., Myers, H. G., Zucker, I. R., Lotman, H., and Asa, M. M., A cardiac pacemaker using biologic energy sources, Transactions on American Society for Artificial Internal Organs, Vol.9, No. (1963), pp.174–177.

Perez, A. J., Leff, R. D., Ip, M. H., and Yang, Z. G, From wearable sensors to smart implants – toward pervasive and personalized healthcare, IEEE Transactions on Biomedical Engineering, Vol.62, No.12 (2015), pp.2750–2762.

Pfenniger, A., Wickramarathna, N. L., Vogel, R., and Koch, M. V., Design and realization of an energy harvester using pulsating arterial pressure, Medical Engineering & Physics Vol. 35, No. (2013), pp.1251–1265.

Sahara, G., Hijikata, W., and Shinshi, T., Implantable power generation system utilizing muscle contractions excited by electrical stimulation, Proceeding of IMechE Part H: Journal of Engineering in Medicine, Vol.23, No.6 (2016), pp.569–578.

Schmidt, C. L and Skarstad, P. M, The future of lithium and lithium-ion battery in implantable medical devices, Journal of Power Sources, Vol.97, No.98 (2001), pp.742–746.

Sohn, W. J., Choi, B. S., and Lee, Y. D., An investigation on piezoelectric energy harvesting for MEMS power sources, Proceeding of IMechE Part C: Journal of Mechanical Engineering Science, Vol.219, No.4 (2005), pp.429–436.

Tashiho, R., Kabei, N., Katayama, K., Tsuboi, F., and Tsuchiya, K., Development of an electrostatic generator for a cardiac pacemaker that harnesses the ventricular wall motion, Journal of The Japanese Society for Artificial Organs, Vol.5, No.2 (2002), pp.239–245.

Tsouri, R. G., and Ostertag, H. M., Patient-specific 12-lead ECG reconstruction from sparse electrodes using independent component analysis, IEEE Journal of Biomedical and Health Informatics, Vol.18, No.2 (2014), pp.476–482.

Venema, B., Gehring, H., Michelsen, I., Blanik, N., and Blazek, V., Robustness, Specificity, and reliability of an in-ear pulse oximetric sensor in surgical patients, IEEE Journal of Biomedical and Health Informatics, Vol.18, No.4 (2014), pp.1178–1185.

Wang, A. D., Chiu, Y. C., and Phan, T. H., Electromagnetic energy harvesting from vibrations induced by Karman vortex street, Mechatronics, Vol.22, No.6 (2012), pp.746–756.

Wu, T., Redoute, M. J., and Yuce, R. M., A wireless implantable sensor design with subcutaneous energy harvesting for long-term IoT healthcare applications, Journal of IEEE Access, Vol.6 (2018), pp.35801–35808.

Zebda, A., Csnier, S., Alcaraz, J. P., Holzinger, M., Goff, L. A., Grondran, C., Boucher, F., Giroud, F., Gorgy, K., Lamraoui, H., and Cinquin, P., Single glucose biofuel cells implanted in rats power electric devices, Scientific Reports Vol.3, Article No. 1516 (2013), pp.1–5.

Zurbuchen, A., Pfenniger, A., Stahel, A., and Stoeck, C. T., Energy harvesting from the beating heart by a mass imbalance oscillation generator, Annals of Biomedical Engineering, Vol.41, No.1 (2013), pp.131–141.