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Efficient wireless power delivery for biomedical implants

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Abstract: Power delivery for biomedical implants is a major consideration in their design for both measurement and stimulation. When performed by a wireless technique, transmission efficiency is critically important not only because of the costs associated with any losses but also because of the nature of those losses, for example, excessive heat can be uncomfortable for the individual involved. In this study, a method and means of wireless power transmission suitable for biomedical implants are both discussed and experimentally evaluated. The procedure initiated is comparable in size and simplicity to those methods already employed; however, some of Tesla's fundamental ideas have been incorporated in order to obtain a significant improvement in efficiency. This study contains a theoretical basis for the approach taken; however, the emphasis here is on practical experimental analysis.

1 Introduction

Implantable biomedical devices have, particularly in recent years, received much attention with regard to their application for a variety of uses involving both stimulation and monitoring. In direct contrast to wearable monitoring healthcare systems [1], one of the key issues for implanted devices is the satisfactory provision of power on an ongoing basis. For short-term experimentation it is quite possible for sufficient power to be provided transcutaneously, a good example of this being the 3-month implantation testing of the Utah Array [2]. For long-term implantation however the power needs can vary considerably [3, 4].

Implants such as the Utah Array have in fact been used for a variety of applications, including an alternative sensory input and neural control of prostheses as well as a new means of communication [5]. Transcutaneous power delivery for any long-term employment will however always carry with it the chance of infection for the recipient as well as the possibility of mechanical leveraging. Clearly an onboard/on chip power pick up device that avoids transcutaneous power supply is an attractive alternative which will most likely result in more widespread use of the technology [6].

Some devices, which require several milliamps of stimulating current, such as those used in the deep brain stimulating electrodes for the treatment of Parkinson's disease, need full battery implantation [7]. This technique then suffers from the requirement of periodic battery replacement. On the other hand, it is possible to consider energy harvesting within the body. This approach is however still in its infancy and its practical usefulness is yet to be fully realised [8].

Wireless power delivery offers the advantages that it reduces the risks (particularly because of infection) associated with either battery replacement surgery or a transcutaneous supply. Inductive coupling can be employed for such power transfer, but the efficiency of transfer is a (very sensitive) function of coil dimensions and the distance between them. Resultant efficiencies for biomedical implants are, as a result, generally very low [9], particularly so in a practical, working environment. The most attractive scheme is arguably therefore coupling between magnetically resonant objects [10].

Although the idea of wireless power has been explored extensively in the literature with several competing power delivery techniques being considered, the most directly relevant are those in which power is not directed, but rather is absorbed [11, 12]. In essence, the required load or draw-down current is determined by the operational constraints and not by the beaming method employed. This feature allows magnetic currents to exist in a passive mode, that is, the energy does not persist in the environment continuously but rather is tapped into on-demand. As a consequence of this, less energy is consumed to drive the circuits.

In this paper we discuss a method for the transmission of power by magnetic resonance in order to power biomedical implants. What is made apparent here is the small size and relatively few components required in the method described as compared to the relatively large and efficient amount of power transmitted.

The transmission frequency selected for the system is 400 kHz. Consideration was given as to whether or not the type of signal deployed, in particular the transmission frequency, had any deleterious effect on human tissue. Clearly this is vitally important if power is to be supplied, such as implants or other wetware, by the method described in this paper. What is important here is the concept of 'safe' zones, that is, signal frequencies for which previous studies have shown that, over a reasonable time period, no mal-effects have been witnessed. According to Soma *et al.* [13] a frequency range of 100 kHz to 4 MHz, at the power levels considered here, has been shown to be suitable for this aim. Our own studies were therefore performed within this frequency range. Nevertheless, in the scheme of magnetic resonance for power transmission, it has already been determined that the transmission does not interact with off-resonant objects [14].

Using an on-board miniature solid-state high-power amplifier, we demonstrate here a prototype capable of delivering reliable operating characteristics suitable for practical implants requiring a steady power supply of anything from 4.5 to 12 V dc.

Contrary to the purposes of other implementations [11, 12, 15–17], of specific interest here, is to question whether a scheme could be accomplished using small coils of a few turns driven by a simple amplification circuit. The goal of this exercise is therefore to establish a driving circuit with a minimum amount of components, thereby reducing device complexity and emitted heat.

As a physical demonstration of the operating characteristics of the method described, we have set a requirement to maintain a sufficiently high-powered signal to reliably power a lamp and motor. The reasoning being that if the technique can function well in terms of such external requirements, it will certainly perform adequately in the case of an implant specification.

In this regard, it has not been our goal to deliver power at the sort of distances reported on in [14, 15, 17] where efficiencies of 40% are perhaps the upper target. Rather, here we attempt to compare directly with transmission over relatively short distances (a few cm), as reported on in [10, 18, 19], with a high efficiency of transmission (over 90%) being our target.

An overriding aim is to overcome power supply issues, eminently apparent in the study of biomedical implants, by realising a wireless scheme which is sufficiently powerful such that an implant can reliably receive its power remotely. Hence, the direction of power transmission as well as the size of technology involved has been important in our studies.

Firstly, in the section that follows, we give a theoretical underpinning of the scheme. Following this we provide some experimental power transmission results. It is worth stressing that these have been obtained from an actual prototype transmission network rather than a computer simulation.

2 Wireless power by magnetic resonance

2.1 Theory of inductive-link power transfer

The concept of wireless power transfer, first described by Poynting [20, 21] and experimentally verified by Tesla [22, 23], has been illustrated in the literature as a viable method to transport electrical current between distant points [10, 18, 24–26]. The extension of this method herein maintains a significant amount of useful power transmitted at intensities of less than 15 V in which, across a volume of air, the magnetic waves are exchanged between two or more coupled resonators.

Consider a circuit consisting of a single loop of insulated wire wound in such a way as to create a circular loop of a few turns. This loop, connected in parallel to a capacitor, becomes one-half of a resonant circuit. This is designated as the transmitter t. The length of the wire loop in t is then replicated to construct a similar loop placed on an independent circuit board positioned a distance away and which is connected in parallel with a capacitor of the same reactive value and a load. In its entirety this becomes the second-half of a resonant circuit, designated the receiver r.

Each half of the resonant circuit is placed a distance from the other. The transmitter is connected to a power source such that the LC circuit is excited. Owing to the symmetry between each half of the circuit, magnetic waves flow from $t \rightarrow r$ at resonance frequency f_0 with an efficiency η. When energised, this circuit engages an electrical field in the loop in t, creating a magnetic field which is coupled to the loop in r. By placing the load in parallel to the loop in r, magnetic energy is converted into electrical current. The arrangement is illustrated in Fig. 1.

A number of receivers can though be fed from one common transmitter, thereby opening up the potential to power mobile robotic platforms by this means. A theoretical configuration of the circuit in a robotic implementation is a single fixed-position transmitter t delivering energy to distant receivers r and s. Such a circuit has been constructed and tested for the intensity of the induced current at different distances away from the transmitter. The receivers are grouped into two categories:

1. AC mode: lighting an incandescent lamp, and
2. DC mode: turning a motor.

Either of the receiver r or s is a lamp or a motor; these become synonymous platforms in a mobile robot. The experimentally tested range of this method is illustrated in Figs. 2 and 3.

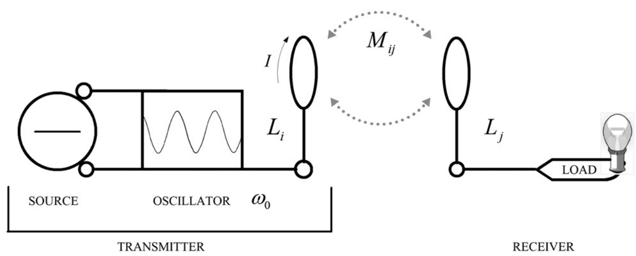

Fig. 1 *Model of inductive-link wireless power*

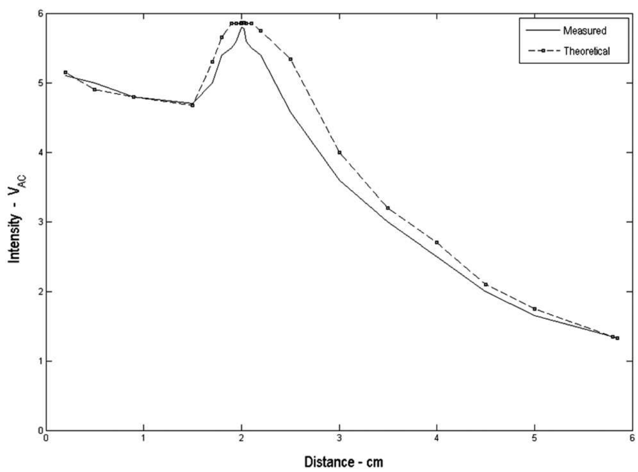

Fig. 2 *AC power intensity over the x–y-plane at a distance with axial alignment*

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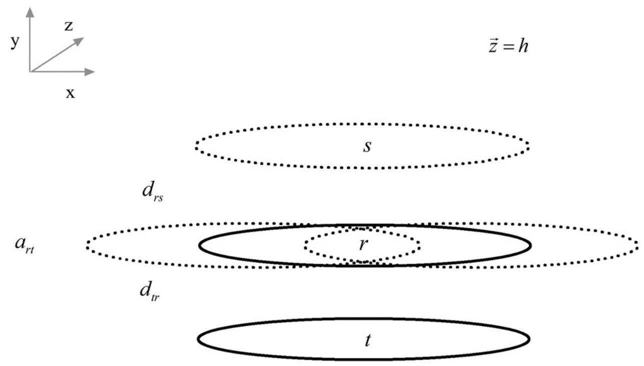

Fig. 3 *Experimentally tested orientation of multiple receivers r and s with reference to each position at the transmitter t when there is no angle of rotation away from axial alignment $a_{rt} = 0$*
Standing height of all coils are equal

2.2 Efficiency of the scheme

The efficiency η of the experimental model is calculated to be [19]

$$\eta = \frac{k_{ij}^2 Q_i Q_j}{1 + k_{ij}^2 Q_i Q_j} \tag{1}$$

where k_{ij} is the coupling coefficient, Q_i, Q_j are the quality factor of the coils L_i and L_j driven at resonance frequency ω_0. Since the efficiency of the given coil geometry is of interest, the coupling coefficient k_{ij}, mutual inductance M_{ij} between distant coils L_i and L_j, and the quality factors of all the coils must be determined.

Meanwhile, the coupling coefficient between magnetically coupled coils in general is defined as

$$k_{ij} = \frac{M_{ij}}{\sqrt{L_i L_j}} \tag{2}$$

where M_{ij} is the mutual inductance, L_i and L_j the self-inductances of the coils.

Further to this the inductance of a circular loop [27] is

$$\begin{aligned} L_i &\simeq \mu_0 \mu_r n_i^2 r_i \left(\ln \frac{8 r_i}{R_i} - 2 + Y \right) \\ L_j &\simeq \mu_0 \mu_r n_j^2 r_j \left(\ln \frac{8 r_j}{R_j} - 2 + Y \right) \end{aligned} \tag{3}$$

where r_i, r_j is the loop radius and R_i, R_j is the wire radius, n_i, n_j is the number of turns and Y is the flow constant of the skin-effect of the emitted radiation.

To determine the mutual inductance M_{ij} apparent in this case, Stokes' theorem [28] yields the value

$$M_{ij} = \frac{\iint_{S_j} \boldsymbol{B} \cdot \boldsymbol{n}\, \mathrm{d}S}{I_{L_i}} \tag{4}$$

where S_j is the area of the receiver coil, $\boldsymbol{B}$ is the magnetic field, I_{L_i} is the current passing through the transmitter secondary coil and $\boldsymbol{n}$ is the vector normal across free space. The loaded qualities, Q_i, Q_j as a property of the capacitance C_i, C_j, resistance R_i, R_j, and inductance L_i, L_j of the circuits containing the transmitter and receiver coils, respectively, are defined [29] as

$$Q_i = \frac{1}{R_i}\sqrt{\frac{L_i}{C_i}}, \quad Q_j = R_j \sqrt{\frac{C_j}{L_j}} \tag{5}$$

The quality factor of the receiver Q_j is shown for the parallel case. For comparative purposes, the quality of the transmitter is also derived from the resonance frequency at the oscillator as

$$Q_i = \frac{\omega_0 L_i}{R_i} \tag{6}$$

When using two AC modes, the quality values in respect of the second receiver are additive to the overall calculation. In this geometry, multiple receivers will exchange magnetic energy with the transmitter and each other. The quality factor, with reference to (5), is based on those properties – inductance, capacitance and resistance – which allow it to manifest a state of resonance between each half of the circuit. By inspection of the equation – and when compared to (6) – the amount of energy available to each receiver is due to the energy consumed in the work being performed given the system's ability to send a limit of quantity of energy given the arrangement within the oscillator. Therefore a system with multiple receivers will have an 'additive' quality in that the transmission will support more than one receiver equally subject to its orientation and position; further, a second receiver (either loaded or unloaded) placed between the transmitter and single receiver, in both AC and DC modes, will allow the amplification of signal over the distance given its ideal position between each, which in the experiments is shown to be one-half the distance. Practically speaking, this is a means of extending the magnitude of the transmission beam over a longer distance by 'chaining' the receivers together.

The resultant theoretically calculated values for the coils are shown in Table 1 and plotted in Fig. 4, which shows both the theoretical and actual experimental results.

2.3 Information transfer

Although not directly examined in this paper, information transfer is also possible by means of the same circuitry, given the sinusoidal nature of the power transmission. Additions to the amplifier circuit would however be required where a modulation signal is introduced to the carrier. Modifications to the receiver would however also be required.

2.4 Experimental conditions

The physical properties of the circuit constructed for experimental purposes are shown in Table 2. The experiment was aimed at answering three questions:

1. What are the suitable geometric positions for the coils in order to deliver power to sufficiently drive a 10 g motor at 6 V, with an operating torque of 3400 g cm?

Table 1 Inductive-link coils theoretical specification

Coil	L, μH	M_{ij}, μH	k_{ij}	Q	C, nF	R, Ω	η	d, cm
L_i	1.57	0.152	0.127	50.48	100	200	81.45	2.05
L_j	1.52	0.149	0.092	5.05	100	20	68.31	2.05
L_k	1.55	0.157	0.099	126.19	100	500	91.47	2.05

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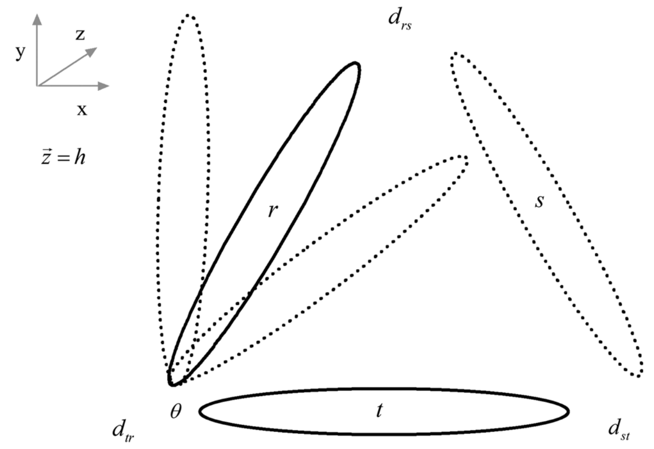

Fig. 4 *Experimentally tested orientation of multiple receivers r and s with reference to each position at the transmitter t with an angle of rotation θ away from axial alignment*

Standing height of all coils are equal

Table 2 Inductive-link coils physical specification

Coil	Coil radius, mm	Wire radius, mm	Wire length, mm	Number of turns
L_i	30	0.4	650	3
L_j	30	0.4	650	3
L_k	30	0.4	650	3

2. What are the suitable geometric positions in order to deliver power to sufficiently drive a 12 W incandescent lamp?
3. What is the consequence in terms of power availability and draw down of adding multiple receivers?

The quantities of measurement required are:

1. Intensity by voltage present in the receiver, in the case of DC mode, from a minimum to maximum position.
2. Intensity by photometric intensity of a lamp, in the case of AC mode, from a minimum to maximum position.

The goal is to calculate the physical properties of the system and see if the practically measured efficiency agrees with the theoretical quantities. To accomplish this we employed two sets of receivers: one pair of lamps and one pair of motors.

3 Experimental results and discussion

Each receiver was operationally tested in one of two modes:

1. AC mode: receiver *r* contains a capacitor and a lamp.
2. DC mode: receiver *s* contains a capacitor, a full-wave rectifier bridge and a DC motor.

Each mode can be used separately or combined.

The receiver circuits in each mode were constructed differently, although each used a coil loop wound with the same physical characteristics as the transmitter coil. Each had a capacitor of the same value placed in parallel.

For one of the AC modes, a bias resistor was added in parallel to increase the *Q* value. For all DC modes, converting the radio-frequency signal to direct current was straightforward. Using 1N34 germanium diodes as a bridge rectifier resulted in minimum voltage drop.

The behaviour of the received current was contrastingly different between each mode. In AC mode, there is a measure of feedback reflected across the circuit, which in this case would initially suggest an improvement of performance, as illustrated in Fig. 4.

However, this comes at a cost. The distance from the transmitter was reduced to 7 cm in AC mode whereas it was held at 10 cm in DC mode.

Using the modes illustrated we next describe here the operational freedom of the mobile receiver.

3.1 AC mode

It can be seen, from Fig. 4, that when using one AC mode, signal performance of a receiver *r* was maximum when its distance d_{tr} from *t* was 2 cm and its axial difference a_{rt} was not greater than 3 cm to the left or to the right of *t*, as was illustrated in Fig. 2. Meanwhile the maximum distance for the receiver when the lamp was absorbing enough power to ignite its filament, was observed at 6 cm.

When using two AC modes (meaning the use of two receivers in AC mode – essentially two lamps operating simultaneously), the signal performance of one receiver *r*, while comparable to one AC mode, consumed power relevant to its proximity with *t*, as expected, but attenuated by a distant receiver *s*. In our experiments the maximum distance when the lamps were absorbing enough power to ignite their filaments was observed at 6 cm. However by lining up *r* and *s* with minimum axial displacement, the currents constructively interfered creating a solenoid structure. This method extended the range of useful induced current to 8 cm, shown in Fig. 5.

When using one AC mode and rotating at an angle θ, as illustrated in Fig. 3, the signal performance of a receiver *r* was observed when θ was between 40° and 90° offset from *t*. In this range, the luminescence falls off and remains steady at $\theta = 90^\circ$.

At a distance d_{tr}, $\theta = 60^\circ$, the lamp was at peak brightness within 1.5 cm, movement along the axial length reduced the signal to its minimum at 3 cm. However, movement along the trajectory of θ reduced the signal to a minimum at 2 cm. Exactly the same behaviour was observed for receiver *s*. The tentative results suggest that the field is consistent across its manifold, that is, consisting of a finite spatial geometry symmetric along its axial length.

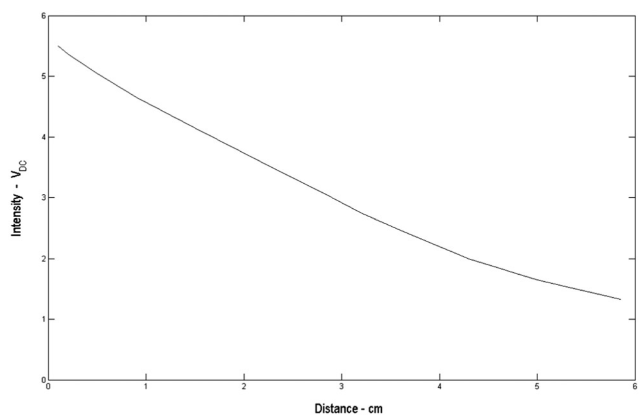

Fig. 5 *Measured DC power intensity over the x–y-plane at a distance with axial alignment*

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When using two AC modes and rotating at an angle θ, the signal performance of the receiver r, although comparable to one AC mode, it was found that for segments of its rotation through θ the energy was exchanged between the two receivers.

When introducing a third and fourth AC mode, for example, adding a third and fourth receiver, the geometric positioning of the intensity of the magnetic field object remained constant and, relevant to the position of the other receivers, more energy was absorbed at the peak, as already described.

3.2 DC mode

When experimenting with one DC mode, using a germanium rectifier bridge, the signal performance of the receiver r was maximum when the distance d_{tr} was at its minimum proximity to t and the axial difference a_{rt} was no greater than 1/3 of the radius to the left or to the right of t, as illustrated in Fig. 2.

In the experiment motor performance was observed at the receiver with sufficient torque to drive the wheel of a robot when the receiving coil was closest to t, with the power induced falling off steadily up to a distance of 4 cm, where it was just sufficient to do the same. When the motor was not loaded, it exhibited a similar performance up to a distance of 9 cm.

The next experiment involved using two DC modes (meaning the use of two receivers in DC mode – essentially two motors operating simultaneously) in a germanium rectifier bridge. The results were found to be comparable to one DC mode in terms of power consumption relevant to its proximity with t, as expected, but the power was attenuated by the effects of the distant receiver s.

The maximum distance, when there was found to be enough power to drive a loaded motor, was observed at 9 cm. By lining up r and s with minimum axial displacement, the currents were made to constructively interfere, creating a solenoid structure. In this way sufficient induced power to drive a motor could be extended to a distance of 12 cm.

When using one DC mode and rotating at an angle θ, as illustrated in Fig. 3, the signal performance of a receiver r was observed when θ was between 40° and 90° offset from t. In this range, the luminescence fell off, remaining steady at $\theta = 90°$.

When using two DC modes and rotating at an angle θ, the signal performance of the receiver r, while comparable to one DC mode, exhibited segments in its rotation through θ where energy was exchanged between the two receivers.

When using one AC and one DC mode, the signal performance of the receivers was found to be comparable in each characteristic performance based on experimental observations. Energy is exchanged so that it is distributed by the position of the receiver relevant to its multiple based on the total energy available in the circuit. By increasing power to the source, higher intensity currents are available to each receiver.

Although demonstrating the transmitter generates magnetic waves which can pass through walls and does not interact with off-resonant objects such as humans or animals, there are reasons where it is undesirable to exchange energy with a circuit which shares the characteristic properties of the resonant circuit. In such cases, shielding of a resonant object is possible by enclosing the machine in thin metallic foil. It was observed that the magnetic waves present in the system cannot penetrate such a substance.

4 System performance

It is possible to affect the Q-factor of the coil by subtracting the series resistance from the transmission circuit, while adding parallel resistance to the receiver circuit. This is achievable particularly when using high-wattage resistors at low quantities of resistance.

For example, by adding a 500 Ω resistor in parallel with the load raises the Q-factor of the circuit thereby causing the lamp to give off a brighter luminescence. In doing so it also improves the efficiency of the circuit. It has to be said however that the addition of resistance into a DC circuit was not tested here.

By increasing the length of the wire after the rectifier circuit, the useful range of the DC signal was extended to 12 cm.

At higher voltages, that is, those at 9 V and greater, the heat given off was found to be minimal in the circuits tested. What heat existed was found to be concentrated around the capacitor in parallel with the loop and was linearly related to the input current.

The maximum power was transmitted in this experiment by driving the amplifier circuit at 12 V. At this level, significant heating of the transmitter's capacitor was found. Changing the input voltage altered the output voltage, that is, there was no apparent storage of magnetic energy in the magnetic field.

Any increase (or decrease) in power supplied does not alter the locations where the coils achieve a maximum or minimum in their resonance coupling, rather it increases (or decreases, respectively) the available power at those locations.

The maximum energy transmitted in our experimentation was 36 kJ. The apparent limiting factor in this scheme was the number of amplifiers we employed. It would be of interest to apply further amplifiers, thereby increasing the power in the system, in order to understand how much current can be practically transported in the geometric space.

Perhaps the most significant results obtained in our experimentation were those relating to the efficiency of power transfer. Here we have aimed to use small, simple and efficient means, at directly comparable distances to those reported in [10, 18, 19]. The main goal of our research programme being to ultimately deliver power for biomedical implants – as described in those papers.

In our case, with two coils, both of radius 3 cm, set 2 cm apart, a measured power transfer efficiency (at a peak) of 91% was achieved. This compares directly and extremely favourably with the 82% reported in [10] and the 80% reported in [19]. It is worth adding that it compares even more favourably with the 45% reported in [18], although it is worth stressing that the goal in that paper was not merely high efficiency.

As an interesting aside, it is also apparent that the same method of power delivery, as tested in our experimentation, can also be employed in the case of multiple receivers. With the addition of additional receivers, the transfer efficiency still remained at a minimum of 82%, while dividing the power – this being based on the position of the receivers relative to the transmitter. Clearly this can mean that the method introduced here could be useful to power several measurement and/or stimulating points within one body, without significant power degradation, from one single external transmission source.

Using two or more receivers, Q will be additive to the solution for each receiver from the perspective of the entire

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efficiency calculation. If interaction 'between' the receivers is considered, modifying (1), given [30], is (see (7)).

If it is desirable to consider the interaction of the magnetic fields of each of the elements: transmitter, receiver one, receiver two etc. operating on each other, then consideration is given that the operation is limited to between a transmitter and receiver one, receiver two 'exclusively'. Therefore the coupling between each receiver is not interesting, rather, noting the improvement in k_{ij} as in (1), then the total, given [29], is (see (8)).

Overall, however, although the separation distance examined here was relatively short (albeit appropriate for biomedical implants), given the radii of the coils, a high amount of power was delivered at high efficiency.

The power issues faced by Barker *et al.* [31] deal with a system consisting of modes – active and sleep – and are concerned with power consumption of a circuitry at random intervals, such as when a vibration triggers the piezoelectric response to generate voltages and currents. Regarding [32], the notion of beacon-delimited, ultra-low body network is comparable to the research herein that they share two common themes: (i) sending power for medical implants or body monitoring apparatus, (ii) sending information at an appreciable data rate; however, the current research only acknowledges that the apparatus has the 'ability' to transmit data given the waveform is a steady-state sinusoid.

5 Comparative analysis

Owing to the importance of the subject area, a body of research, both theoretical and experimental, has appeared, particularly in the last decade, on this topic. As a result we have attempted to make it clear in the paper how our work compares favourably with that of others in terms of the results obtained. It is important however to also give an indication of how these improvements in the results have been achieved through original differences in fundamental design.

It must be realised that the basic ideas underlying the differences have been sparked by returning to some of the fundamental work of Tesla [22] and Poynting [20]. As such it is made clear in this section that the geometry of the magnetically coupled coils we employed was not 'standard' but rather exhibited differences to that used in other work, for example, [10, 18, 19, 33]. Here we point out some of the differences.

Firstly, we made use of a Litz winding, not wound in any particular orientation. This same approach was in fact used in [10], but not so in [18, 19, 33]. Secondly, the radius of the receiver and transmitter coils in our experimentation was the same (3 cm), others meanwhile used one radius for the transmitter and another for the receiver [10–12, 19, 33]. On top of this we used very few turns on both coils ($n = 3$) and did not use a secondary coil. This is in direct contrast to both [10, 18].

Also diverging with previous work, we used an analog oscillator, integrated with the loops, that is, the inductor and capacitor formed a resonant circuit. We also used plain, circular loops, whereas [4, 9, 19] all used flat coils whereas in [10] the authors used a staggered arrangement. In this way our loops could 'consume' the entire AC cycle. We also used a lower resonant frequency, that is, longest wavelength, than that in previous research.

Overall however the key to our approach was simplicity. We appear to have used, by quite some way, the fewest number of components (13 in total for the oscillator) and yet generated a clean sinusoidal waveform at 400 kHz. On top of this our receiver contained only three components. By contrast, in [33] the authors employed a noisy digital driver which also required supplemental circuitry.

6 Concluding remarks

In this paper, the method and means of effective wireless power for the purpose of powering biomedical implants has been described. The focus of this paper is on the description and experimental testing of power delivery for such an application area.

However, the method of power delivery is general and can be applied to other applications which require wireless power transfer on scales both large and small [14, 25].

The experimental results show that significant improvements in terms of power transfer efficiency are achieved by directly connecting the LC circuit to an amplifier circuit instead of excitation being achieved from an external sinusoidal source.

Our measured results were found to be in very good agreement with the theoretical models obtained separately and match well with the simulation results. Our next step clearly will need to involve tests involving actual biological tissue. Although, by analysing the results from previous comparable studies [10, 19], it is not anticipated that there will be any issues of significance, nevertheless such a study is a necessity before the actual practical application of the procedure can go ahead in situ.

In the tests it was found that the efficiency of the energy transfer system can be improved by increasing the Q-factor of the coils. In this way the prototype power transfer system experimented with here achieves at least 4× more efficiency and power density in watts per centimetre, given its small size, compared to prior inductive-link schemes.

The reasons behind the improved results, as reported in this paper, in comparison with the previous work, are not particularly because of the antenna type or Q-factor of the coils, as one might expect. Rather, we feel it is a property of the oscillator driving the system under resonance. In our design we have used a 'magnetic' loop which has an advantage when its circumference is at a ratio of the driving frequency (quarter) wavelength. The oscillator frequency has then been set so the loop can be small and non-toxic in human presence. It is the manner of how the driving frequency is realised, how clean the oscillator is in synchronising voltage and current in phase, that determines the quality of the energy exhibited in the receiver.

$$\eta_{\text{Total}} = \frac{(k_{ij}^2 Q_i Q_j)(k_{jk}^2 Q_j Q_k)(k_{kl}^2 Q_k Q_l)}{[(1 + k_{ij}^2 Q_i Q_j)(1 + k_{kl}^2 Q_k Q_l) + k_{jk}^2 Q_j Q_k][1 + k_{jk}^2 Q_j Q_k + k_{kl}^2 Q_k Q_l]} \tag{7}$$

$$\eta = \frac{1}{[1 + (1/k_{ij}^2)((1/Q_i) + (1/R_i))((1/Q_j) + (1/R_j))((1/Q_k) + (1/R_k))][(1 + (R_j/Q_j))(1 + (R_k/Q_k))]} \tag{8}$$

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As reported in the previous section, efficiency results of 91% in power transfer over a range of 2 cm between transmitter and receiver coils are extremely encouraging. Our next step is to extend the length of transmission to over 7 cm in AC mode by the addition of a secondary coil whose winding is equidistant on the plane. For biomedical implants this then starts to look at the possibility of remote wireless power provision, in which case the opportunity for multiple recipients also becomes relevant.

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