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<td>Author(s)</td>
<td>Yeung, Sai Ho; Pradhan, Raunaq; Feng, Xiaohua; Zheng, Yuanjin</td>
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Comparing the magnetic resonant coupling radiofrequency stimulation to the traditional approaches: Ex-vivo tissue voltage measurement and electromagnetic simulation analysis

Sai Ho Yeung, Raunaq Pradhan, Xiaohua Feng, and Yuanjin Zheng

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Comparing the magnetic resonant coupling radiofrequency stimulation to the traditional approaches: Ex-vivo tissue voltage measurement and electromagnetic simulation analysis

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Recently, the design concept of magnetic resonant coupling has been adapted to electromagnetic therapy applications such as non-invasive radiofrequency (RF) stimulation. This technique can significantly increase the electric field radiated from the magnetic coil at the stimulation target, and hence enhancing the current flowing through the nerve, thus enabling stimulation. In this paper, the developed magnetic resonant coupling (MRC) stimulation, magnetic stimulation (MS) and transcutaneous electrical nerve stimulation (TENS) are compared. The differences between the MRC RF stimulation and other techniques are presented in terms of the operating mechanism, ex-vivo tissue voltage measurement and electromagnetic simulation analysis. The ex-vivo tissue voltage measurement experiment is performed on the compared devices based on measuring the voltage induced by electromagnetic induction at the tissue. The focusing effect, $E$ field and voltage induced across the tissue, and the attenuation due to the increase of separation between the coil and the target are analyzed. The electromagnetic stimulation will also be performed to obtain the electric field and magnetic field distribution around the biological medium. The electric field intensity is proportional to the induced current and the magnetic field is corresponding to the electromagnetic induction across the biological medium. The comparison between the MRC RF stimulator and the MS and TENS devices revealed that the MRC RF stimulator has several advantages over the others for the applications of inducing current in the biological medium for stimulation purposes. © 2015 Author(s). All article content, except where otherwise noted, is licensed under a Creative Commons Attribution 3.0 Unported License. [http://dx.doi.org/10.1063/1.4930533]

I. INTRODUCTION

The application of magnetic resonant coupling (MRC)\(^1\) has been very popular in wireless power transfer due to its high efficiency. Meanwhile, the coil design using MRC has recently been proposed for electromagnetic therapy applications\(^2\) such as non-invasive radiofrequency (RF) stimulation. The setup consists of two coils, namely primary coil and secondary coil, coupled to each other. Both coils are cascaded with resonating networks, which are designed so that both coils are resonating at same frequency. Although there are two coils, only one active source is used to drive the primary coil. The secondary coil does not need another active source, and it is driven by the magnetic field coupled from the primary coil. At resonance, the primary and the secondary coil are conducting the same current. Both current conducting the primary coil and the secondary coil would superpose the magnetic field contributed by them around the biological medium such as tissue placed in between the two coils. This technique is beneficial to generate a higher magnitude of electric field $E$ at the stimulation target compared to the case when a single coil is used. The higher the magnitude of the $E$ field, the higher the current passing through the biological medium, leading to a more effective neural stimulation and hence, MRC is a very promising technique for neural stimulation using MRC. However, from existing related studies, there is lack of a comparative
study of MRC and traditional electric or magnetic based neural stimulation techniques which are magnetic stimulation (MS)\(^3\)–\(^{11}\) and transcutaneous electrical nerve stimulation (TENS).\(^{12}\)–\(^{14}\) In these existing techniques, MS applies a magnetic coil, which is conducting a high magnitude of current with a low frequency of 1 – 10 kHz, to induce electric field in the biological medium which stimulates the nerve at the target. Meanwhile, TENS attach two electrodes at the two parts of human body, where current is conducted between the electrodes and passes through the human tissue that serves the stimulation effect.

In this paper, the RF stimulation technique using MRC is compared to MS and TENS. The comparisons are based on the operating mechanisms, ex-vivo tissue voltage measurement and electromagnetic simulation analysis. The comparisons in the paper will show that the MRC RF stimulator has advantages over MS and TENS. Firstly, from the electromagnetic theory point of view, the electric field intensity generated by electromagnetic induction from magnetic coil is proportional to the operating frequency. It follows that MS, which is operating at a much lower frequency, needs a much higher current conducting the coil to induce the same magnitude of current density at the biological medium compared to the MRC RF stimulation. Secondly, from the ex-vivo tissue voltage measurement, although both the MRC RF stimulator and MS stimulator induced voltage inside the tissue, the attenuation due to a longer distance from the coils become less in the MRC RF stimulator compared to the MS. Thirdly, from the electromagnetic simulation, the MRC RF stimulator achieved good focusing effect which focuses the electric field at the center. Furthermore, TENS has poor deep penetration and localization effect compared to MRC RF. Therefore, the MRC RF stimulator is a better device compared to the traditional MS and TENS in terms of inducing current in the biological medium for stimulation purposes. The above mentioned comparisons are presented in detail within the following sections.

II. MECHANISM OF OPERATION

This section will introduce the mechanism of operation of MRC RF Stimulation, MS, and TENS.

A. Magnetic Resonant coupling (MRC) Radiofrequency (RF) stimulation

The MRC RF stimulation applies a pair of coils, coupled by magnetic resonant coupling, to induce strong RF electromagnetic field and hence current on human nerve for stimulation purpose. In the literature, RF wave has been applied widely in pulsed electromagnetic field therapies.\(^{15}\)–\(^{17}\) The system diagram of the MRC RF stimulation is shown in Fig. 1. The signal generator generates the signal for stimulation which is amplified by a pulse power amplifier. Two resonant networks resonate with the primary and the secondary figure-of-eight coils respectively, so that they resonate at the same RF frequency for the condition of MRC. A matching network is then used for impedance matching to the MRC resonator. Under MRC, the primary coil conducts the same magnitude of current as the secondary coil. Both currents then induce RF electric field to the stimulation area, thereby inducing magnetic field in the biological medium placed in between the coils according to the Biot-Savart law.\(^{18}\) The magnetic flux density \(\mathbf{B}\) generated by a circular coil of \(N\) turns with negligible thickness can be computed by integrating along the whole current path \(C\) of the coil:\(^2\)

\[
\mathbf{B}(\mathbf{r}) = \frac{\mu_0 N I}{4\pi} \oint_C \frac{d\mathbf{l}' \times (\mathbf{r} - \mathbf{r}')}{|\mathbf{r} - \mathbf{r}'|^3}
\]  

where \(\mu_0\) is the permeability, \(N\) is the number of turn of the coil, \(I\) is the magnitude of current flowing through the coil, \(d\mathbf{l}'\) is the differential coil element, \(\mathbf{r}\) and \(\mathbf{r}'\) are the position vectors of the observation point and that of the differential element \(d\mathbf{l}'\) respectively. Since the current \(I\) is changing in the pulse waveform, the magnetic flux density \(\mathbf{B}\) also changes with time. Then, the change of magnetic field would induce electric field in the biological medium of stimulation, governed by the Faraday’s law:\(^{18}\)

\[
\nabla \times \mathbf{E} = -\frac{\partial \mathbf{B}}{\partial t}
\]
With the electric field induced, the current density $J$ in the biological medium during stimulation can be computed by:

$$ J = \sigma E $$

where $\sigma$ is the conductivity. The current flowing through the nerve results in stimulation effect.

When a human body part is placed in between the primary and the secondary coils, current would be induced at the nerve which initiates the stimulation process. A burst of RF pulse signal is then presented as a waveform from the stimulation process. The pulse operation with a low duty cycle ($\leq 1\%$) ensures the power dissipation in the biological medium is low which ensures RF safety is taken into consideration.

The waveform of the MRC RF stimulator is given as:

$$ f(t) = A \sin 2\pi f_c t, \mod(t, T_r) < T_d $$

$$ f(t) = 0, \text{ otherwise} $$

where $t$ is the time, $T_d$ is the duration of the RF pulse, $T_r$ is the repetition period of the RF pulse, and $f_c$ is the frequency of the RF pulse. For the actual implementation of the MRC RF stimulator, $f(t)$ is applied as a voltage signal, which is generated by a signal generator and amplified by a power amplifier. $A$ is the voltage magnitude from the signal generator of the MRC RF stimulator, which can be amplified through a power amplifier. The MRC stimulator voltage waveform $f(t)$ is proportional to the magnitude of current conducting the coil $I$ and hence proportional to the magnitude flux density $B$ in the differential equation (2). The proportional relation between the magnitude of current conducting the coil and the magnitude flux can be observed in equation (1). The larger the magnitude of the voltage $f(t)$, the larger the current $I$, and hence the larger the magnetic flux density $B$.

Then, $\mod(t, T_r)$ is modulus after $t$ is divided by $T_r$. The duty cycle $D$ is given by:

$$ D = \frac{T_r}{T_d} $$

Using a small duty cycle, it can decrease the average power dissipated to the tissue by multiplying the peak power to the duty cycle. The peak power $P_P$ and the average power $P_A$ are related by the duty cycle $D$ as:

$$ P_A = P_P \times D $$
For instance, assuming a large peak power $P_p$ of 100 W is applied with the duty cycle $D$ of 1%, the average power would become 1 W.

The advantage of applying RF stimulation instead of low frequency is that the electric field increases with the increase in the conducting current of the coil. To demonstrate this effect, the electric field $E$ generated by a circular coil of $N$ turns with negligible thickness can be computed by integrating along the whole current path $C$ of the coil:

$$E(r) = \frac{\mu_0 f_c N I}{2} \oint_C \frac{dl'}{|r - r'|}$$

where $f_c$ is the RF frequency. It can be easily concluded that the higher the frequency, the higher the magnitude of the $E$ field; hence a larger current is thus induced in the biological medium as given in equation (3) due to its proportional relationship with current density. Although some previous analysis\textsuperscript{20} shows that the stimulation threshold in also increases with the frequency, so that the rate of increase of the threshold would be different to the rate of increase of the $E$ field intensity. For instance, when the stimulus frequency increases (in a continuous sine wave), the rate of the increase in reaction thresholds of daily cows is slower than the rate of increase in stimulus frequency.\textsuperscript{20} In any case, the operating frequency of the MRC RF system is adjustable by changing the capacitance value of the capacitor in cascade with the coil because

$$f_0 = \frac{1}{2\pi \sqrt{LC}}$$

\textbf{B. Magnetic Stimulation (MS)}

MS conducts a high current pulse to a magnetic coil and induces electric field in the biological medium which stimulates the nerve at the target. Two types of pulses are common in MS, which are monophasic pulse and biphasic pulse. Unlike the MRC (where RF waveform is applied for stimulation), MS uses a monophasic or biphasic pulse for stimulation. The frequency used in MS is 1-10 kHz, so the pulse duration is about hundreds of microseconds.\textsuperscript{11}

The biphasic pulse\textsuperscript{3} is considered in this paper since it is more effective and the MS system architecture is depicted in Fig. 2. It usually consist of a magnetic stimulation driver which charges up and discharge a capacitor\textsuperscript{3} to generate a pulse. The pulse passes through a magnetic coil and generates magnetic field across the target biological tissue medium according to Biot-Savart law (Equation (1)). The magnetic field then induces electromotive force on the biological tissue and thus inducing current based on Faraday’s law.

A sinusoidal biphasic pulse can be represented by:

$$f(t) = A \sin 2\pi f_c t, \text{ mod}(t, T_r) < 1/f_c$$
$$f(t) = 0, \text{ otherwise}$$

where $t$ is the time, $T_r$ is the repetition period of the biphasic pulse, and $f_c$ is the frequency of the biphasic pulse. The biphasic waveform can be generated by charging up a capacitor, followed by

\begin{figure}[h]
\centering
\includegraphics[width=0.5\textwidth]{figure2.png}
\caption{Magnetic stimulation system architecture.}
\end{figure}
discharging it so that the current stored in the capacitor flow to the magnetic coil to generate a large magnetic field for electromagnetic induction at the target biological medium.

C. Transcutaneous Electrical Nerve Stimulation (TENS)

TENS applies two electrodes at two parts of human body which conducts a current between them. The current passes through the human tissue which serves the stimulation effects and in comparison to the MRC and RF stimulation, TENS lacks localization effect and deep penetration because the skin is not a good conductor and has resistance which affects the current reaching the nerve when the voltage is applied at the surface of the skin. The TENS system architecture is depicted in Fig. 3. The current flowing $I$ from the TENS device into the surface of the skin can be computed by the Ohm’s law:

$$I = \frac{V_S}{R_e}$$

where $V_S$ represents the voltage generated by the TENS device and $R_e$ is the equivalent total resistance between the positive and the negative electrodes. Meanwhile, there are different layers of the skin and the current flows in multipath from one electrode to another. The current across the stimulation area of interest would be less than the total current $I$ and more current is expected to flow near the surface of the skin instead of deep penetration because the path is shorter which results in less resistance.

III. EX-VIVO VOLTAGE MEASUREMENT ON TISSUE

In this section, ex-vivo tissue experiments of voltage measurement are done by measuring the voltage across a box of pig muscle tissue (i.e. pork cut) and the size of the tissue is selected based on the spatial resolution of the focal region, which is dependent mainly on the radius of the magnetic coil.

A. Focal region of the magnetic coil and the size of the tissue

The magnetic coil used for MRC RF stimulation has 20 turns with a radius of 9 mm. The large radius and the large number of turns were chosen because these two parameters are both proportional to the magnitude of $E$ field induced at the focal region for each ampere of current conducting the coil due to equation (7), and hence capable of generating a larger current across the biological medium due to equation (3). To visualize the focal region, the $E$ field distribution along the three Cartesian axes is computed as shown in Table I, where the $E$ field distribution along the axes are plotted in Fig. 4. The field magnitude is computed by equation (7), while also taking the phase difference between the two coils (i.e. 65°) into consideration. In equation (7), the integral can be computed by integrating every differential element of the coil, where each differential element is divided by the absolute distance between the differential element and the observation point. Since the MRC RF stimulator consists of the primary coil and the secondary coil, the $E$ field contributed by both coils should be computed and added together. The focal region is defined by half power

![FIG. 3. Transcutaneous electrical nerve stimulation system architecture.](image-url)
TABLE I. The calculated \( E \) field distribution per ampere conducting the coils at different location along the three Cartesian axes.

| Location along the axis (cm) | Electric Field Intensity \(|E|\) (V/m) |
|-----------------------------|--------------------------------------|
|                            | \( x \)-axis | \( y \)-axis | \( z \)-axis |
| -15                         | 3.19        | 1.39        | 2.14        |
| -14                         | 3.74        | 1.65        | 2.65        |
| -13                         | 4.32        | 1.97        | 3.31        |
| -12                         | 4.87        | 2.37        | 4.21        |
| -11                         | 5.24        | 2.88        | 5.43        |
| -10                         | 5.20        | 3.51        | 7.12        |
| -9                          | 4.51        | 4.31        | 9.53        |
| -8                          | 3.06        | 5.30        | 13.06       |
| -7                          | 0.93        | 6.51        | 18.48       |
| -6                          | 1.70        | 7.94        | 27.45       |
| -5                          | 4.62        | 9.55        | 43.99       |
| -4                          | 7.63        | 11.23       | 44.29       |
| -3                          | 10.50       | 12.80       | 28.44       |
| -2                          | 12.95       | 14.10       | 20.42       |
| -1                          | 14.63       | 14.94       | 16.45       |
| 0                           | 15.23       | 15.23       | 15.23       |
| 1                           | 14.63       | 14.94       | 16.45       |
| 2                           | 12.95       | 14.10       | 20.42       |
| 3                           | 10.50       | 12.80       | 28.44       |
| 4                           | 7.63        | 11.23       | 44.29       |
| 5                           | 4.62        | 9.55        | 43.99       |
| 6                           | 1.70        | 7.94        | 27.45       |
| 7                           | 0.93        | 6.51        | 18.48       |
| 8                           | 3.06        | 5.30        | 13.06       |
| 9                           | 4.51        | 4.31        | 9.53        |
| 10                          | 5.20        | 3.51        | 7.12        |
| 11                          | 5.24        | 2.88        | 5.43        |
| 12                          | 4.87        | 2.37        | 4.21        |
| 13                          | 4.32        | 1.97        | 3.31        |
| 14                          | 3.74        | 1.65        | 2.65        |
| 15                          | 3.19        | 1.39        | 2.14        |

FIG. 4. The calculated \( E \) field distribution per ampere conducting the coils at different location along the three Cartesian axes.
change from the central point of the MRC coupled coil ($x = y = z = 0$ cm), which is given by (i) $x$-axis: -2.9 cm to 2.9 cm, $y$-axis: -4.3 cm to 4.3 cm, $z$-axis: -2.2 cm to 2.2 cm. Therefore, the dimensions of the tissue used in ex-vivo experiment should be smaller than the focal region. The size of the tissue is chosen as 2.5 cm ($x$-axis) × 3.5 cm ($y$-axis) × 1.0 cm ($z$-axis), whereas the voltage is excited by different stimulators including MRC RF, MS, and TENS. The box of tissue for ex-vivo measurement is depicted in Fig. 5. The piece of tissue is stored in a plastic box, where aluminum electrodes of dimensions approximately 1.0 cm × 2.5 cm are attached to the two sides of the piece of tissue. The voltage of the piece of tissue would then be measured through the edges of the aluminum electrodes using an oscilloscope with wire connections.

B. Magnetic Resonant coupling (MRC) Radiofrequency (RF) stimulation

The ex-vivo voltage measurement setup using MRC RF is given in Fig. 6. The piece of tissue is placed in between two figure-of-eight coils. One coil is driven by a pulse power amplifier and a signal generator as previously depicted in Fig. 1, while another coil is driven by MRC. The radius of the coils is 4.5 cm with 20 windings at each side of each figure-of-eight coil while the distance between the two coils is 9 cm. Since the piece of tissue is placed at the center between the two coils, the distance between the piece of tissue and either coil is 4.5 cm.

Due to Biot-Savart law and Faraday law as given in equations (1) and (2), electric field and current would be induced across the piece of tissue. Thus, voltage is induced across the two edges of the piece of tissue that can be measured. In the experiment, a 450.5 kHz RF signal is used to drive the MRC coil and hence, the overall MRC RF stimulation system induces a transient response where the induced voltage start from the zero value to the steady state value. RF voltage pulse duration of 1 ms, which is measured from the tissue placed at the middle between the two coils is
given in Fig. 7 and the voltage magnitude becomes steady at around 0.15 ms. Since the measured magnitude of voltage is scalable with the source signal magnitude as well as the distance between the tissue and the coils, the absolute magnitude of the waveform is not the primary interest but rather the shape and transient response of the waveform.

The transient response at startup of the measured magnitude of voltage is also shown in Fig. 8(a). To show the effect of the attenuation due to the distance between the coil and stimulation point, the voltage measured at steady state when the tissue is placed at different location in the z-axis is shown in Fig. 8(b). It should be noticed that the transient response would be some different when the tissue is placed at different locations because the transient response of the current conducting the two coils is also different.\(^3\) However, the transient response should not be the main concern
FIG. 8. Measured voltage response of the RF MRC stimulation across the piece of tissue.

(a) The transient response of the measured voltage when the tissue is placed at the center

(b) The measured voltage at steady state at different location along the z-axis

as Fig. 7 shows that the main portion of measured waveform is in steady state. Meanwhile, the transient response of the current $I$ can be taken into account in the calculation of the field in equations (1) and (7). However, it should be noticed that the transient response is not the main concern when the waveform duration is significantly long compared to the transient response duration.

Since the primary and the secondary coils are placed at -4.5 cm and 4.5 cm, respectively, the measured voltage is stronger when the tissue is placed near the either coil. From the figure, it can be seen the trend of attenuation of electric field intensity along z-axis that was given in Fig. 4 is very similar to that of the voltage induced, which is due to the proportional relationship between the $E$ field and the voltage:

$$V = d |E|$$

where $d$ is the distance between the two terminals of the voltage measurement, and the assumption is constant $|E|$ magnitude between the two terminals. The conductivity of the tissue has certain effect on the magnitude of $E$ field, which can be observed from the following time harmonic equation
derived from the ampere law:\textsuperscript{18}

\[
E = \frac{1}{\sigma + j\omega\varepsilon} \nabla \times H
\]  

(12)

where \(\omega\) is the angular frequency and \(\varepsilon\) is the permittivity. From equation (12), it is expected that the magnitude of \(E\) field is much smaller in tissue (e.g. muscle tissue assuming \(\sigma = 0.8 \text{ S/m}\)) compared to air (\(\sigma \approx 0 \text{ S/m}\)) at 450 kHz. From the equation, the magnitude of the induced \(E\) field becomes different in different medium, where the change of magnitude is reflected in the denominator term of equation (12).

C. Magnetic Stimulation (MS)

The ex-vivo voltage measurement setup using a commercial magnetic stimulator (Model: MagVenture MagPro Compact\textsuperscript{®}) is given in Fig. 9. The magnetic stimulator generates a 280 \(\mu\)s biphasic pulse across the figure-of-eight coil (Model: MagVenture MC-B35). The surface of the coil is attached tightly to the piece of tissue. According to the specification of the figure-of-eight coil model, it has an inner radius of 24 mm and an outer radius of 47 mm, winding height of 9 mm, and number of winding given by \(2 \times (3 \times 4)\).

Similar to the MRC RF stimulation, MS generate magnetic field from current conducting coil across the stimulation region according to Biot-Savart law, and the changing magnetic field induces electric field in the piece of tissue and hence current due to electromagnetic induction (Faraday’s law). The voltage is then induced at the edge of the piece of tissue and measured as shown in Fig. 10. In the measurement, the tissue is placed 2.5 cm away from the coil. Again, the magnitude of voltage induced is scalable by changing the input source level and the distance between the coil and the piece of tissue, and hence the waveform is the primary interest. The biphasic pulse contains both the positive and negative voltage in the total timespan of approximately 280 \(\mu\)s. When the tissue is located at different distances from the coil, the corresponding maximum voltages of the pulses measured across the tissue are shown in Fig. 11.

FIG. 9. Ex-vivo voltage measurement on pig muscle tissue with the magnetic stimulator.
D. Transcutaneous Electrical Nerve Stimulation (TENS)

The ex-vivo voltage measurement setup using a commercial transcutaneous electrical nerve stimulator (Model: Genial® JOZ-B62) is given in Fig. 12. The stimulator generates a biphasic voltage pulse train across two electrodes which is attached at the edges of the piece of tissue, thereby conducting pulse current to the piece of tissue. The voltage across the piece of tissue is then measured as shown in Fig. 13. The TENS device generates a biphasic pulse train in a total duration of 15.6 ms and becomes idle for another 17 ms before the start of the next pulse train. Again, the induced voltage could be adjusted by changing the voltage applied to the electrodes.

IV. ELECTROMAGNETIC SIMULATION OF ELECTRIC AND MAGNETIC FIELD DISTRIBUTION

In the previous section, we have shown how the stimulation waveform could be obtained from the MRC stimulator, commercial MS and TENS devices. However, computing the electric field and current distributions of the tissue is quite difficult; therefore, we shall make use of electromagnetic
stimulation with the aid of CST Studio Suite®. In the simulation, a rectangular piece of tissue would be assumed, so that the result would be comparable to the previous section concerning measurement. Although the frequency response of the tissue often changes with frequency, the fixed tissue parameters of $\sigma = 0.8$ S/m and $\varepsilon_r = 56.8$ will be used in the following analysis.
The electric field distribution is directly related to the current distribution in the tissue due to the proportional relation in equation (3).

A. Magnetic Resonant coupling (MRC) Radiofrequency (RF) stimulation

MRC figure-of-eight coils as well as a piece of tissue are modelled in the electromagnetic simulation as shown in the left subfigure of Fig. 14. In the model, current of the frequency 450 kHz flows through the coils and generates magnetic field which induce current in the tissue. The distribution of the electric field $E$ in the tissue when the size are $2.5 \text{ cm} \times 3.5 \text{ cm} \times 1.0 \text{ cm}$ and $20 \text{ cm} \times 10 \text{ cm} \times 5 \text{ cm}$ are shown in Figs. 15(a) and 16(a), respectively. In the figures, the brighter the colour, the larger the magnitude of the electric field. It can be seen that the electric field intensity, which is proportional to current density, is strongest at the center for both tissue sizes. Hence, it is shown that the figure-of-eight MRC coil for RF stimulation focus its stimulation effect at the center of the tissue. Therefore, the electromagnetic simulation result gives similar characteristic to the calculated result in Fig. 4, where the focusing location is also at the center. Then, the distribution of magnetic flux density $B$ as demonstrated in Fig. 17(a) shows that the magnetic field penetrates the tissue and is essential for electromagnetic induction to generate current in the tissue. Unlike the electric field where the strongest magnitude occurs at the center, the magnetic field is weaker at the center than the locations around it. However, the magnitude of $B$ at the center does not contribute to the current density at the same location as indicated in equation (3), where magnitude of $E$ is the point of emphasis.

B. Magnetic Stimulation (MS)

A figure-of-eight coil as well as a piece of tissue are modelled in the electromagnetic simulation as shown in the middle subfigure of Fig. 14. Current of frequency 10 kHz flows from the electrical source to the coil which induces magnetic field. The magnetic field radiated from the coil induces current in the tissue due to electromagnetic induction. The distribution of the electric field $E$ in the tissue when the size are $2.5 \text{ cm} \times 3.5 \text{ cm} \times 1.0 \text{ cm}$ and $20 \text{ cm} \times 10 \text{ cm} \times 5 \text{ cm}$ are shown in Figs. 15(b) and 16(b), respectively. The electric field and therefore current is induced in the tissue in both tissue sizes, but the distribution appears to be different from the MRC RF stimulator. Meanwhile, just as in the case of the MRC RF simulator, the magnitude of the $E$ field is stronger at the center than of the locations around it, which is due to the focusing effect achieved by the figure-of-eight coil. Then, the distribution of magnetic flux density $B$ is shown in Fig. 17(b), which also demonstrates that the magnetic field can penetrate the tissue which is essential for electromagnetic induction. Again, the magnitude of $B$ at the center does not contribute to the current density at the same location, but the magnitude of $E$ is proportional to the current density as indicated in equation (3).
C. Transcutaneous Electrical Nerve Stimulation (TENS)

The connecting wires from the TENS source as well as a piece of tissue are modelled in the electromagnetic simulation as shown in the right subfigure of Fig. 14. In the model, voltage applied to the electrodes and the current flows directly from the TENS sources into the tissue. Since from...
the literature research in Ref. 8 (Table II), the transcutaneous neural stimulation with the frequency ranging from 14 Hz to 360 Hz leads to excellent pain relief, the frequency applied in this paper is in the mid-range 100 Hz. The distribution of the electric field $E$ in the tissue when the size are $2.5 \, \text{cm} \times 3.5 \, \text{cm} \times 1.0 \, \text{cm}$ and $20 \, \text{cm} \times 10 \, \text{cm} \times 5 \, \text{cm}$ are shown in Figs. 15(c) and 16(c), respectively. It can be seen that the electric field intensity, which is proportional to current density, is strongest near the electrodes and the central shortest current path for both tissue sizes and the
electric field intensity attenuates far from them. It is because more current would flow through the shortest path between the two electrodes which results in less resistance compared with other longer paths. This implies that the TENS lacks deep penetration and localization effect. The distribution of magnetic flux density $B$ is shown in Fig. 17(c) and although the source of magnetic field is not the coil, the magnetic field is generated by the current conducting the connecting wires, electrodes, and tissue due to the Maxwell-Ampere law: \[ \nabla \times \mathbf{H} = \mathbf{J} + \frac{\partial \mathbf{D}}{\partial t} \] (13)
V. CONCLUSION

The MRC RF stimulation is compared to the traditional MS and TENS in terms of the mechanism of operation, ex-vivo tissue experiments, and electromagnetic simulation. Since the electric field intensity generated by electromagnetic induction from magnetic coil is proportional to the operating frequency according to equation (7), the electric field generated from the low frequency MS (1 – 10 kHz) should be lower (at least 45 times lower) than that generated by the MRC RF stimulator (450 kHz) based on the same current. Since electric field is proportional to the current density generated at the biological medium, it follows that the MS needs a much higher current (at least 45 times higher) to induce the same magnitude of current density compared to the MRC RF stimulation.

From the ex-vivo tissue experiments, both the MRC RF stimulator and MS stimulator induced similar amount of voltage in the tissue. However, the attenuation due to a longer distance from the coils becomes less pronounced from the MRC RF stimulator when compared to the MS. In the literature of coil design consideration, the ratio of the stimulation intensity observed at two different depths represents the attenuation in depth. Hence, the attenuation due to a longer distance from the coils in the MRC RF stimulator as well as that of the MS can be observed by the following figure-of-metric $M$:

$$M = \frac{V_{\text{focus}}}{V_{1.0\text{cm}}}$$

where $V_{1.0\text{cm}}$ and $V_{\text{focus}}$ are the voltage when the tissue is placed at 1.0 cm away from coil and at the focus respectively. The focus is defined by the center between the two MRC coils. The comparison of the figure-of-metric is based on the two facts: firstly, the longer the distance from the coil, the larger the attenuation of $E$ field and hence, lower voltage is detected. Secondly, different coils have a different attenuation rate based on the distance which is reflected by the value of $M$. The values of $M$ of the MRC RF stimulator and the MS, captured from Figs. 8 and 13, are 0.5514 and 0.1693 respectively. This shows that the MRC RF stimulator attenuation is lower compared to that of the MS. This is contributed by the fact that the MRC RF stimulator is operated by the magnetic field produced from the two coils rather than only one coil in the case of MS. Both coils would contribute to the electric field and superposition at the center, thereby reducing the spatial attenuation from the coils.

From the electromagnetic stimulation, all the approaches induced electric field in the tissue which is proportional to the current density due to equation (3). The MRC RF stimulator achieved good focusing effect which focuses the electric field at the center as observed from Figs. 15 and 16. It results that the current density is larger at the center compared to that of the locations around it. Therefore, relating to the real RF stimulation application, it facilitates the stimulation taking place at the target instead of stimulating various locations at the same time in human body. Also, the distribution of magnetic flux density $B$ as shown in Fig. 17 demonstrates that the magnetic field penetrates the tissue which facilities electromagnetic induction to generate current through the tissue.

Meanwhile, for the case of TENS device, the electric field intensity is strongest near the electrodes and the central shortest current path, but attenuates greatly at the locations far from them. This shows that the TENS lacks deep penetration and localization effects compared to the MRC RF stimulator.

The differences between MRC RF stimulation and MS in the simulation results of the electric field and magnetic field distribution are due to the coil configuration (MRC figure-of-eight coils versus a single figure-of-eight coil) as well as the difference in the frequency of operation (450 kHz versus 10 kHz). Meanwhile, the electric field distribution of the TENS is totally different from MRC RF stimulation and MS because the current is conducted directly through electrodes of the TENS devices and the electric field is determined by the current conducted as indicated in equation (3).

Finally, it is worthy of note that there are three advantages for the MRC RF stimulator. Firstly, MRC RF stimulator requires a lower current conducting the coils to induce the same magnitude of current density compared to the MS. Secondly, MRC RF stimulator attenuation is lower with respect to the distance compared to the MS. Thirdly, the MRC RF stimulator has better focusing performance compared to the TENS devices.
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