Declaration of Compliance for the Ohio Willow Wood second generation Limb Logic Vacuum System with inductive charger

Overview

Ohio Willow Wood is submitting final documentation for our inductively charged prosthetic suspension system called the Limb Logic Vacuum Suspension system or LLVS. The LLVS is used to suspend prosthetic limbs on amputee patients. The system is charged with an inductive charger comprised of control electronics, a ferrite core primary winding located in the charger plug, and a receiving coil located in the product to be charged. (Figure 1 and Figure 2)

Prior to submission of FCC compliance documents to the Curtis Straus TCB, Ohio Willow Wood submitted test and modeling data to the OET in compliance with the requirements of notice "680106 D01 RF Exposure Wireless Charging Apps v01." This submittal is located in KDB 880208 ranging from 10/2011 to 11/2012. After significant assistance from the OET, Ohio Willow Wood received notification of compliance to the OET's intended requirements for inductive charging on 11/15/2012.

The KDB process resulted in a number of alternate designs and test reports. This document is intended to summarize the relevant portions of the discussion / documentation process in KDB 880208. For a complete record of the transaction, please refer to the KDB transcript.

Compliance is demonstrated in this report by modeling after confirming model accuracy with field measurements near the tip of the charger plug, a region of the field that is relatively uniform. Sensor and field measures used to verify the Finite Element Model (FEM) data were witnessed at Curtis Straus, a Bureau Veritas company, located at One Distribution Center Circle, Suite #1, Littleton, MA 01460. All modeling was performed at Ohio Willow Wood using QuickField modeling software made by Tera Analysis Ltd located at Knasterhovvej 21, DK-5700 Svendborg, Denmark.

This document contains five sections: Section One is a functional description of the charger, load circuitry, and their function, Section Two is a description of the sensor used for the evaluation and the methods used to verify calibration, Section Three is a description of the Finite Element Model (FEM) and a comparison of modeled field data along the Z axis of the primary charger winding to measured values, Section Four discusses the rational agreed to by OWWCO and the OET and lists relevant limits for SAR and E fields based on FCC and ICNIRP guidelines, and Section Five is an analysis of the model data and a comparison to the limits described in Section Four.

Section 1: System description.

The G2 inductive charger design is based on a pair of coils used to form an inductive link between the charger electronics and the device to be charged. The primary coil is wound around a ferrite rod and is driven by an H-Bridge design at one of two fundamental frequencies: 84 kHz or 42 kHz. This coil is then potted into a plastic enclosure referred to as the plug. (Figure 1) At 84 kHz, current through the primary winding is limited by the inductance of the primary winding to levels below the ICNIRP recommended limits. When the primary is plugged into a compliant load, the load detects the presence of a charger and then toggles a load across the secondary coil on and off at either 250 Hz or 350 Hz depending on whether the load is charged or charging. (Figure 2) This modulation in the impedance of the load is detected by the drive electronics of the charger as a "tone" and the charger controller then shifts the drive frequency to 42 kHz to allow higher power to be coupled to the load. When the plug is removed, the charger electronics detect the loss of the low frequency tone and reverts to 84 kHz.

Figure 1
Charger "plug"

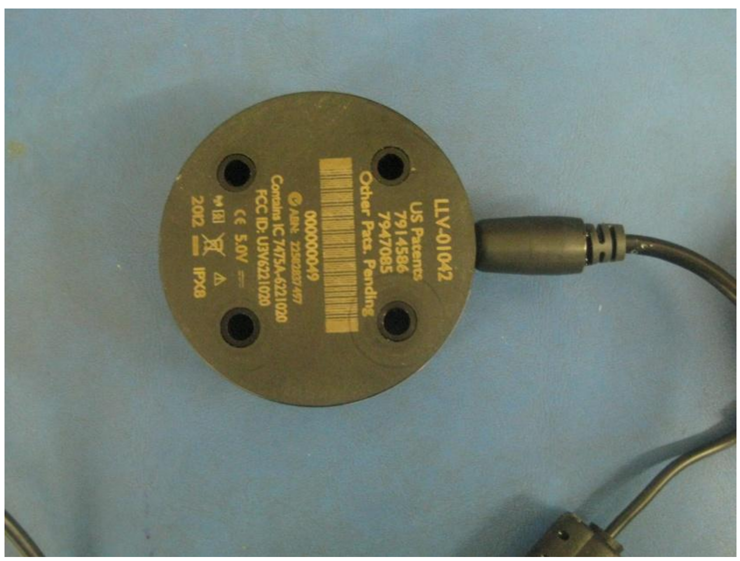

Figure 2
Charger "plug" inserted into load

The charger (schematic 6226010) is powered by an external, 13.8 Vdc, medical grade supply (attached at JP1) and its output is plugged into a load (schematic 6226060001). The charger circuitry includes regulated supplies at 12.4 and 3.3 volts. (U4, U1 respectively) An H-Bridge is powered through a Schottky diode from the nominal 13.8 volt supply. This supply has a tolerance of ±5% and therefore a maximum voltage of 14.49 Vdc. This Schottky diode prevents reverse biases from the inductive load from impacting on the medical supply and other circuitry and has a forward voltage of 0.25 Vdc around the operating point of the charger. The resulting voltage driving the H-Bridge is nominally 13.55 Vdc but could range up to 14.24Vdc based on the tolerance of the supply. For this reason, the model was validated at 13.55 Vdc, but compliance was based on simulations at 14.24 Vdc.

In normal function, the charger starts up in 84 kHz mode and watches for a tone. Once a tone is detected, indicating that the plug is plugged into a valid device, the charger shifts to 42 kHz mode. The charger shifts back to 84 kHz mode within 250 ms of losing the tone.

Exposure of the user to 42 kHz would require that they either 1) pull the plug out of the unit under charge and then place it in contact with tissue in < 250ms, or 2) experience a double fault of the charger. In the 42 kHz band, the unit operates between the ICNIRP and IEEE limits. However, even at 42 kHz, the E fields generated in tissue only exceed the ICNIRP general public criteria by 19% and falls below the occupational exposure limits. Further, this exposure is only possible within several mm of the plug. For these reasons, the likelihood of harm due to a system failure is highly unlikely.

The leaky I^2T integration was added because, while the unit complies with the ICNIRP and IEEE limits, it was determined that the unit could still drive sufficient power in the 84 kHz mode to heat up a single turn of a fine wire wrapped around the charger tip to the point that it would cause discomfort to the touch. In the current design, a single turn is detected and shut off in <250 ms, a duration that was found to:

1) Not heat up a single turn of #30 AWG wire placed tightly around the charger tip above temperatures comfortable to the touch.
2) Allow the charger to pull enough power to start sending back tones.

In the event that the unit detects a load but no tone, it shuts down for 3 seconds, and then retries at 84 kHz, sufficient time for a shorted single turn to cool.

In a normal charge sequence:

1) The charger starts up and the H-Bridge drives the coil at 84 kHz.
2) The charger tip is placed into the device to be charged.
3) The voltage across C111 in the device to be charged builds to >3.0 volts at which point the device to be charged detects the chargers presence.
4) The device to be charged begins to toggle R38 on and off modulating the impedance of the load.
5) The impedance of the load is processed in the microprocessor using a pair of Goertzel filters to look for the presence of either a 250 or 350 Hz tone.
6) When the tone is detected, T5 in the charger circuitry is turned on increasing the capacitance in the RC timer circuit for the H-Bridge driver and dropping the H-Bridge frequency to 42 kHz.
7) The battery charger is enabled by the load.
8) The charger remains in 42 kHz mode as long as there is a valid low frequency tone detected.

Section 2: Sensors and Calibration

As suggested by the ICNIRP documents, we used FEA modeling to validate our design. One of the challenges faced was that the dimensions of the primary core are on the order of 3mm and the field is highly variable over distances much larger than this size. Since there are no commercially available probes of this size calibrated for 42 kHz, validating the model was difficult. To do so, a small probe was constructed and calibrated in a long solenoid at 105 kHz and 42 kHz. The probe consisted of 20 turns of #40 AWG single insulated magnet wire wrapped on a 3.1mm diameter plastic rod. Calibration was performed by construction of a large air core solenoid. The solenoid was constructed with 138 turns of #20 AWG single insulated magnet wire on a 21.3 mm plastic pipe. This construction resulted in a length to diameter ratio of 5.95 indicating a very stable B field inside the solenoid. The solenoid was driven with a sinusoidal voltage at a number of amplitudes and a range of frequencies. Pictures of these windings are shown in **Figure 3.**

Figure 3
Sensor and Calibration Windings

To calibrate the sensor, the solenoid was first calibrated. To do so, the solenoid was driven with a function generator through a 1 ohm resistor. This allowed both the solenoid voltage and current to be measured. This in turn allowed the inductance of the solenoid, allowing for its resistance, to be compared to the theoretical inductance of:

$$L = \frac{Uo\ A\ N^2}{\ell}$$

Because the B field in a solenoid is:

$$B = UoNI/\ell$$

where:

B the magnetic field strength
N is the number of turns
I is the current and,

ℓ is the length of the coil

and:

$$\phi = BA$$

where:

ϕ is the magnetic flux
A is the cross sectional area of the coil

and inductance (L) is defined as

$$L = N\phi/I$$

It can be shown that

$$L = NBA/I$$

Or that the inductance of a solenoid is proportional to the B field inside the solenoid.

Using this relationship, the variation between L and the theoretical value for L was used to scale the calculated B field in the center of the solenoid at a given frequency. The sensor was then introduced to the center of the solenoid and the voltage generated into a 1 MOhm input impedance oscilloscope probe was measured as a function of both field amplitude and field frequency. Probe gains were found to be quite stable at a given frequency and were recorded at a number of drive voltages for a range of frequencies including 84 kHz and 42 kHz.

Section 3: Finite Element Model and validation

The charger primary is constructed with turns of heavy insulated magnet wire on a 1/8" diameter ferrite rod with a length of 1 inch. The windings are placed in multiple layers starting 1/8" from one of the ends of the rod.

Any modeling scheme makes assumptions. The following is a list of those made in documenting conformity for the OET.

1) Our software assumes that a material is either conductive or insulating. This results in the use of only conductivity or permittivity. Calculated frequency based adjustments to the complex impedance resulting from permittivities are smaller than the margins allowed for by assuming conductivity of 1S/m. This is a standard assumption in all of the modeling papers pointed to by the ICNIRP documents.

2) $E = J / \sigma$. Largely follows from 1. Stated as a basic assumption in all of the ICNIRP documents and supporting literature.

3) E fields measured at any reasonable tissue conductivity in tissue are going to be the same. Modeling at $\sigma = 1.00$ and 0.05 S/m resulted in magnitude changes in the E field of <0.1%. This indicates that the resulting J fields are small enough to not have a significant back EMF that could affect the overall field thus indicating that the calculated E field would be valid across the range of tissue conductivities.

4) In air or free space, we set the conductivity to both 1e-6 and 1e-9 S/m and then apply E = J / σ. Resulting E field data from these two conditions was the same to within 1%.

   Again, the modeling software that we are using tends to focus on either magnetic fields, conductivities and current; or electric fields, voltages, and permittivities. Crossing over requires indirect measures. The internal math is still solid and the models are consistent with reality when measured.

5) We use an axis symmetric model with the axis running direction down the center of the main coil. Conductivities on the main axis are not allowed by the software when using circuit simulations because they tend to result in singularities in the calculations. To address this we placed a 0.0005 inch gap between all conductive materials and the axis. This corresponds to a 0.001" hole running down the center of the core and the conductive media. Field modeling resulted in no differences in the fields except directly at this "hole" in our model so we have assumed that this does not affect the results for the overall model.

6) Our charger drives the coil with a square wave with an amplitude of 13.8 Vdc. While the modeling software will model square waved simulations, in practice, individual runs with our model take weeks. The sinusoidal simulations only take a few hours. The ICNIRP documents allow square wave data to be compared if the peak of the square wave is modeled to be the peak of the sine wave. While this logic follows for stimulation where peak E is the issue, it misses the point for SAR values. For SARs, we modeled using sine waves with RMS values equal to the peak value for our square wave. This also provides an additional 41% margin to the stimulation values while bringing the heating model into line with the additional power available from the square wave.

7) Permeability for the core was set at a fixed relative permeability of 77. This is the nominal on the core and we operate at a linear region of the BH curve <1/2 B sat.

The model is shown in **Figure 4**. It consists of the same number of turns of the same size conductor as the plug design wound on a ferrite rod of the same average permeability (relative permeability of 77) and conductivity (1 S/m) as the rod used in the design. This is surrounded by a shell (permeability 1, conductivity 0) with dimensions matching the enclosure that the primary winding is potted into during the manufacturing process. To verify this model, it was simulated in air (permeability of 1, conductivity of 0) with the coil driven by 13.55 Vdc at both 105 kHz and 42 kHz, the voltage measured on the driving electronics at the coil.

Final operating frequencies were later changed to address other concerns of the OET. However, these values still bracketed the final operating range and demonstrate the accuracy of the model over the test range.

Model derived values for the B field along the Z axis were compared to values measured with the sensor described in section 2. The resulting data, both simulated and collected, are plotted in

Figure 5. The two data sets correlate well. The standard deviation of the errors is 4%. The maximum error seen for any sample in the data set was 10%.

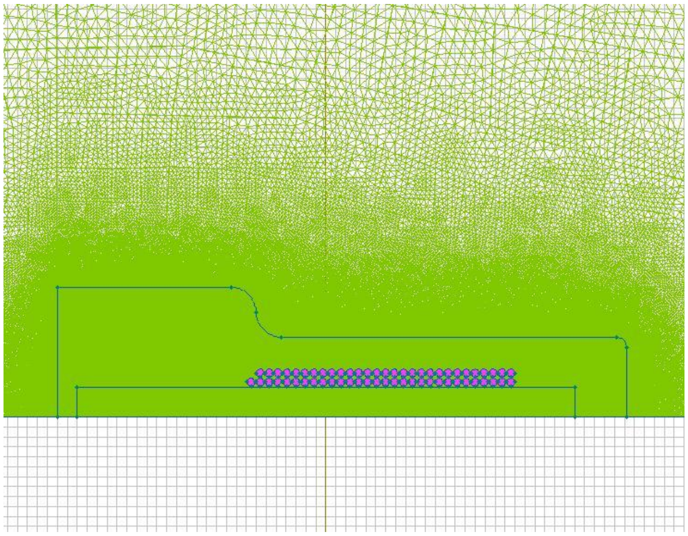

Figure 4
FEM model of charging primary

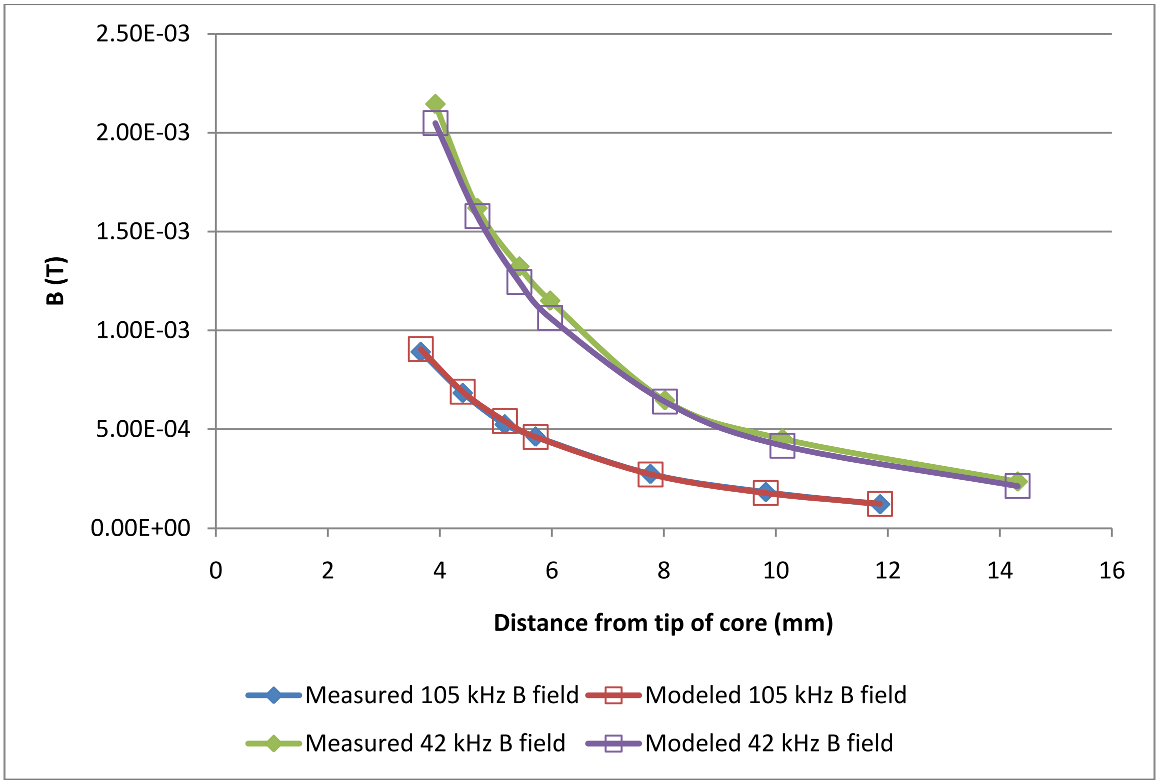

Figure 5
Measured and simulated data used to validate the FEM

Section 4: Compliance rational

EM fields primarily interact with tissue in two ways. The first is by generating local eddy currents that generate ohmic heating resulting in temperature increase in the tissue. This effect is addressed by Specific Absorption Rate (SAR) limits. For the purpose of this evaluation, the FCC proposed a limit of 1.6 W/kg for any 1 gram element. The second effect is the result of interactions of the local E field and nervous tissues. If the field is sufficient, current is driven through the membranes of local nerves and an action potential results in stimulation of the nerve. The 2010 ICNIRP guidelines provide Basic Restrictions for human exposure to time-varying electric and magnetic fields as they relate to this potential to stimulate nervous tissue. The document suggests that modeling be used to compare specific device performances to these values to evaluate compliance.[i] This document supersedes a 1998 document[ii] that originally covered both the stimulation of nervous tissues and field absorption leading to tissue heating. The first paragraph of the 2010 document states that:

> "For the purpose of this document, the low-frequency range extends from 1 Hz to 100 kHz."

Based on this introduction to the 2010 ICNIRP document, and agreement with the OET, the voltage based Basic Restrictions in Table 2 of the 2010 document are used to measure compliance with limitations related to the stimulation of nervous tissue.

The ICNIRP document provides reference levels to allow easy verification of larger or lower power systems by taking measurements in the air around the unit to be tested. However, this document also states that:

> "If measured values are higher than reference levels, it does not necessarily follow that the basic restrictions have been exceeded, **but a more detailed analysis is necessary to assess compliance with the basic restrictions."**

Because direct measures are at best difficult, if not impossible, on a device of this size, Ohio Willow Wood chose to procure modeling software and to then construct and validate a model. In this way, it is possible to directly assess compliance with the Basic Restrictions of the two documents.

The limits of both documents allow for general public limits and occupational limits. We have assumed that the most restrictive of these, public exposure, is appropriate. This sets the bar for nervous stimulation limits at a potential field of

42kHz

$$\mathbf{Vmax(RMS) = 1.35 \times 10^{-4} \times 42{,}000 = 5.7\ V/m,\ or}$$

$$\mathbf{Vmax(peak) = \sqrt{2} \times 1.35 \times 10^{-4} \times 42{,}000 = 8.2\ V/m.}$$

150kHz

$$\mathbf{Vmax(RMS) = 1.35 \times 10^{-4} \times 79{,}000 = 10.7\ V/m,\ or}$$

$$\mathbf{Vmax(peak) = \sqrt{2} \times 1.35 \times 10^{-4} \times 79{,}000 = 15.1\ V/m.}$$

Unfortunately, the ICNIRP Basic restrictions are stated as a single RMS value evaluated at a single frequency. The H-Bridge drives a square wave field. To bridge this gap, we relied on the ICNIRP's reference document on evaluating non-sinusoidal waveforms.[iii] This document states acceptable level of equivalence of a pulse of a given width to a sinusoid with a period of twice this width and an amplitude equal to the amplitude of the pulse. Our square wave is in essence a pulse train and is evaluated as such. This simplification of the problem allows a direct comparison of the peak values in our modeled data to the peak values computed earlier.

The last challenge is the limitation of the modeling software to not reporting the E field. To account for this, we rely on setting a reasonable conductivity for the tissue and application of a base assumption that:

$$J = \sigma \times E.$$

This, in turn, yields:

$$E = J / \sigma.$$

Where J is the current density in A/m^2, σ is the conductivity in Siemens/m, and E is the potential field in V/m.

As might be expected, the conductivity of different tissues varies significantly. Reported values for conductivity at 100 kHz vary from approximately 0.05 S/m for adipose tissue to as high as about 1 S/m for wet skin.[iv] For this reason, simulations were run for both of these values and peak readings were taken to demonstrate sensitivity to conductivity and compliance to the limits stated above.

It should be pointed out that the 2010 ICNIRP document suggests cubical elements with 2mm sides and the FCC standards request 1g which corresponds to 1cc of tissue or elements of approximately 10mm on a side. Our software uses an adaptive meshing system that resulted in elements much smaller than either of these. However, peak values were below the values derived above. As such, spatial averages could not exceed the ICNIRP E filed limits or FCC SAR levels.

Section 5: Modeling Data

Based on the agreement found between the model and measured values in air, we then placed the modeled plug into a modeled uniform conductance media and measured the induced currents. Potential fields were calculated as J/σ and these E fields were compared to the limits derived in section 4.

Evaluation of effects of tissue conductivity

To bound the effects of tissue conductivity on the resulting E field, currents were measured under two conditions: 84 kHz and 105 kHz with direct tissue contact.

Color plots of the J fields for 42 kHz function and 105 kHz function in both a 0.05 S/m and a 1.00 S/m environment are shown in Figure 6 - Figure 9. If one considers that the height of the flat rectangular section of the enclosure on the right is only 4mm high, it can be appreciated that the field only extends a few mm from the plug. Maximum values observed in the model outside of the plastic enclosure are shown in Table 1.

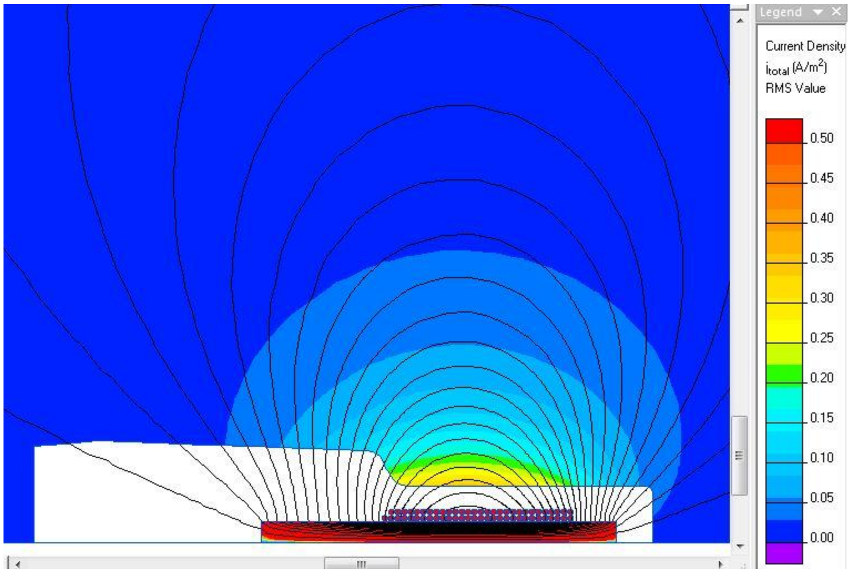

Figure 6
42 kHz into 0.05 S/m

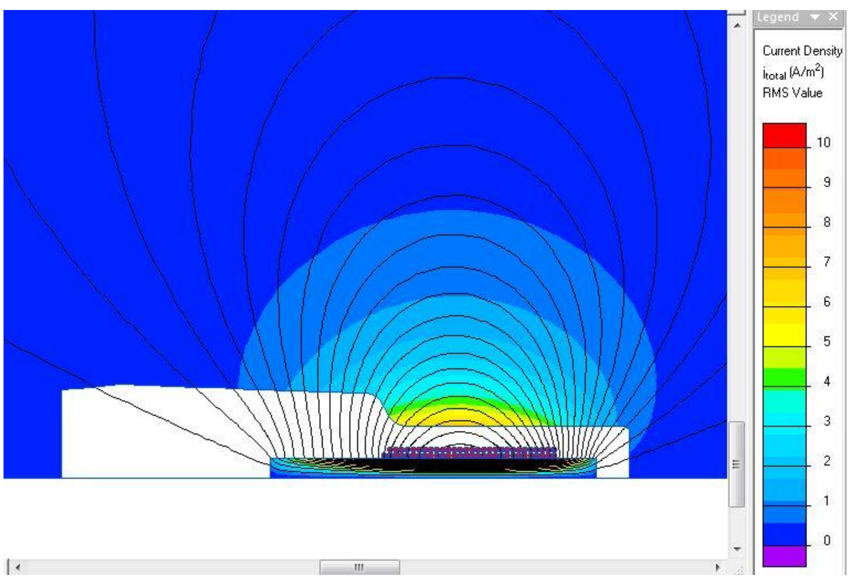

Figure 7
42 kHz into 1 S/m

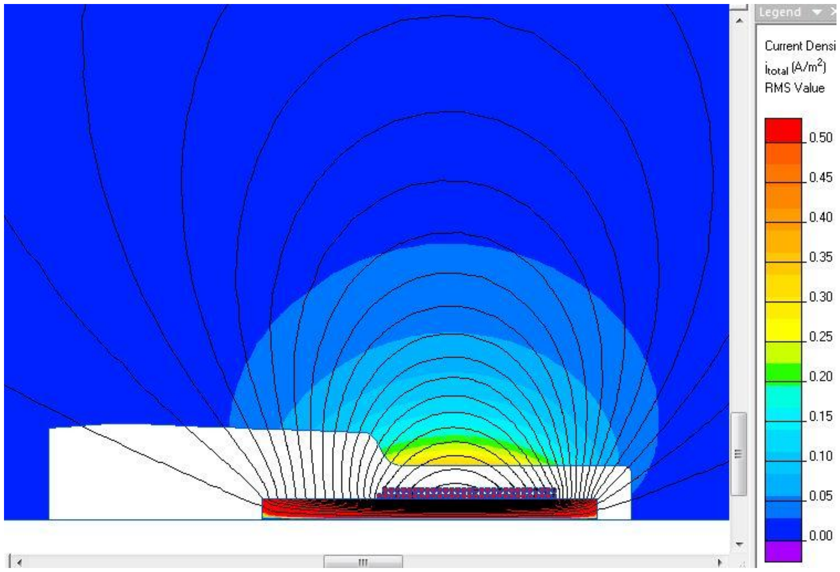

Figure 8
105 kHz into 0.05 S/m

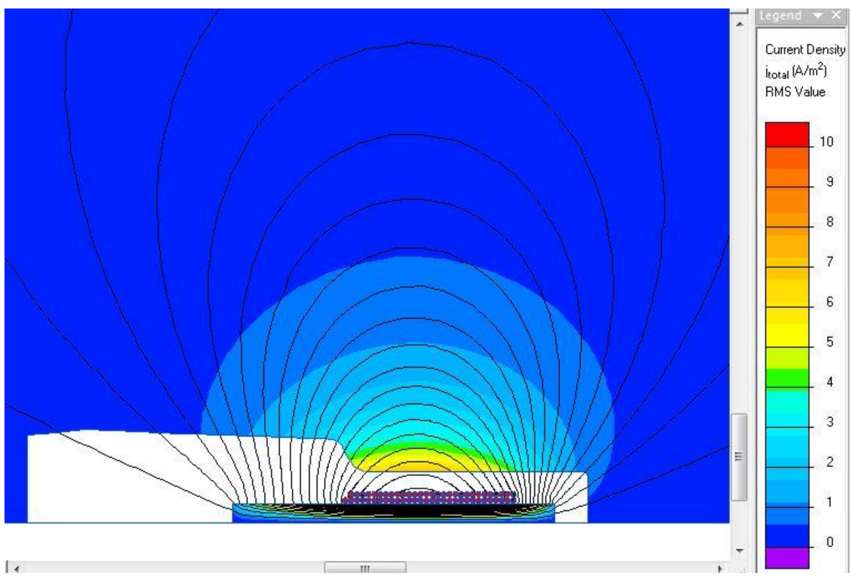

Figure 9
105 kHz into 1.0 S/m

Frequency (kHz)	J 0.05 (S/m) (A/m^2)	E 0.05 (S/m) (V/m)	J 1 (S/m) (A/m^2)	E 1 (S/m) (V/m)	E limit (V/m)	Power 0.05 (S/m) (W/kg)	Power 1 (S/m) (W/kg)	Limit (W/kg)
105	0.487	9.74	9.75	9.75	20	0.00475	0.0951	1.6
42	0.488	9.77	9.74	9.77	8.2	0.00477	0.0950	1.6

Table 1
Effects of tissue conductivity

As can be seen in Table 1, changing the conductivity of the media by a factor of 20, a range that covers the range of expected tissues conductivities, had little effect on the resulting E fields.

E fields at 42 kHz exceeded the limitations set by the ICNIRP by 19% and this was the driver for the two frequency system. Because this band of relatively high induced E field is limited to within several mm of the sides of the plug, because it is only generated when the plug is inserted into a verified load, and because the primary and secondary are in a closely coupled condition in this case, the 42 kHz signal does not pose a risk to the user as will be shown in later documentation.

SAR data

The FCC limit of 1.6 W/kg evaluates to 1600 W/m^3. Compliance with this limit was verified at 79 kHz, the worst case tolerance stack up frequency, in direct contact with tissue, and at 42 kHz with the additional spacing provided by the receptacle. Values were further evaluated at 42 kHz with several receiver side loads that bounded the expected range of power draws in a functioning system.

42 kHz SAR data

To account for fields in the charging, low frequency state, we needed to account for the load imposed on the plug when it is plugged into a receiver and load. To do so, we created an accurate model of our receiving coil with the same number of turns of the same gage wire with the same spacing and dimensions as a production receiver coil. After working with the provider of our modeling software, we were also able to include a model for a linear load for this coil. We were not able to simulate a bridge rectifier or the other circuitry in our system. For these reasons, this evaluation is based on a model of a simple resistive load across the receiving coil that approximates the RMS load imposed on the receiving coil by a system under charge. We believe that the modeling shows that a load has no significant effect on the user's exposure.

Models were driven with 13.8 *1.414 = 19.51 V amplitude sinusoidal waveforms to simulate similar RMS heating effects of our 13.8 amplitude square wave driver signal. The section on assumptions discusses this in more depth.

For modeling purposes, we measured the current and voltage delivered from production receiving coils to receiving units under maximum charge rate conditions. The measured voltage was 7.65 VAC at 0.220 mA. This worked out to an effective impedance of 34.7 Ohms. When the receiving coil was shunted directly into a 10 kOhm load, the lowest load that will be seen when plugged into a receiver with the system off and a fully charged battery, the measured voltage was 9.3 VAC. Based on these measured values, we used 34.7 Ohms and 10 kOhms as limits for the model and added in a data point at 69.4 Ohms in the middle to verify that the effect was essentially monotonic. Because the 34.7 Ohm model ignores the nonlinear effects of the bridge and includes all other assumptions in the model, it is a rougher approximation than other data presented so far. However, we believe that the margins observed in the SAR and ICNIRP data, and the trends seen when adding in a load, justify using the modeled values for the liner load.

The modeled load drew 270 ma at 9.3 volts. This deviates from the measured values by 21%, but this probably reasonable considering the compounding of errors and assumptions. When the model is run with a straight 10 kOhm load, the modeled output voltage was 9.6 VAC. This is a closer match to measured values, about 3% off, and we believe that it indicates that much of the 21 % is due to unaccounted for nonlinear properties of the load during charging.

Modeling results for SAR's at 42 kHz when plugged into a load are shown in Figure 10
42 KHz 34.7 Ohm load SAR dataFigure 11, and Figure 12. The field lines in Figure 10 were accidently scaled differently than in Figures Figure 11 and Figure 12. As can be seen in the heating component of the plots, there was little actual difference in the field intensities.

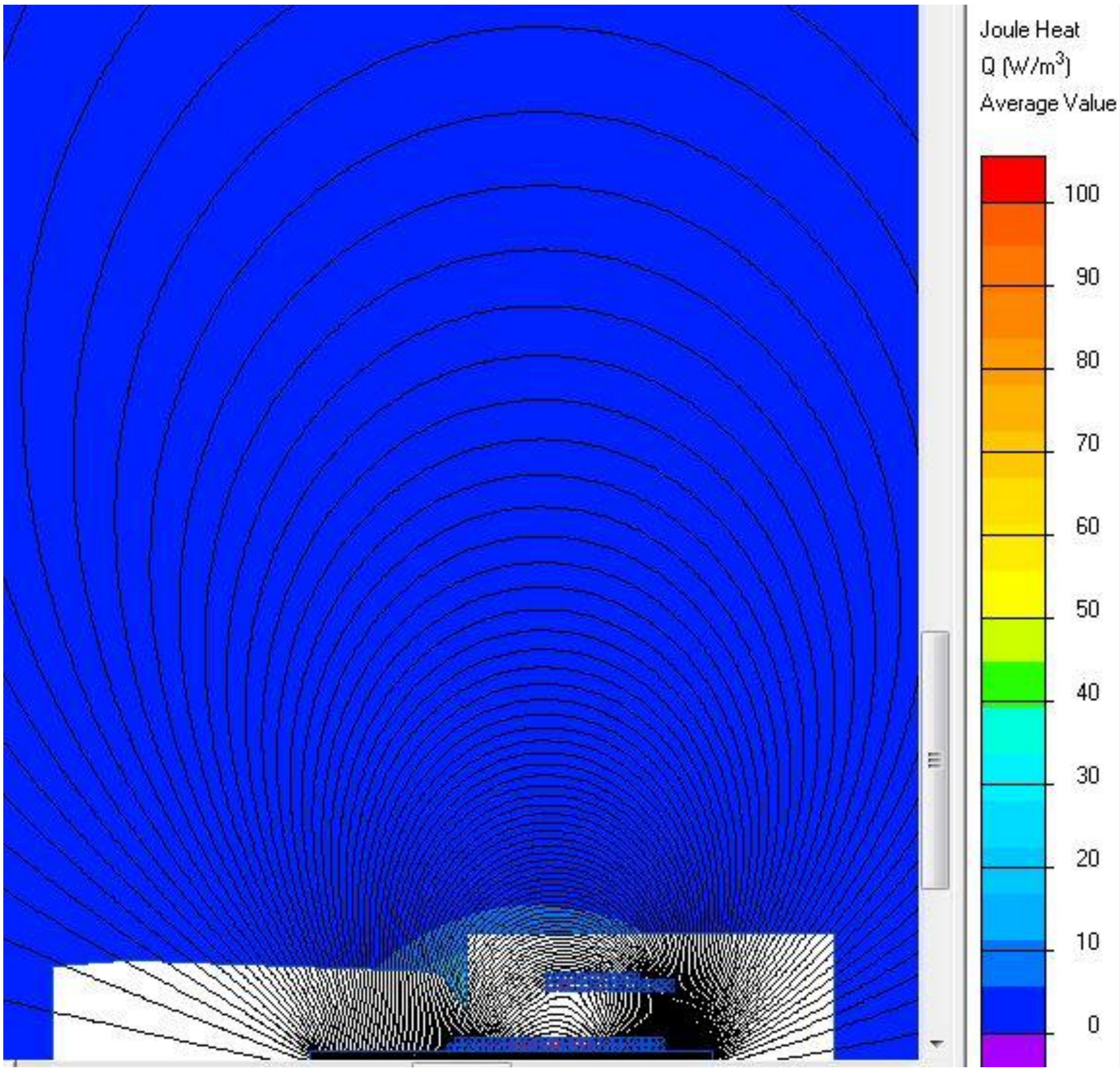

Figure 10
42 KHz 34.7 Ohm load SAR data

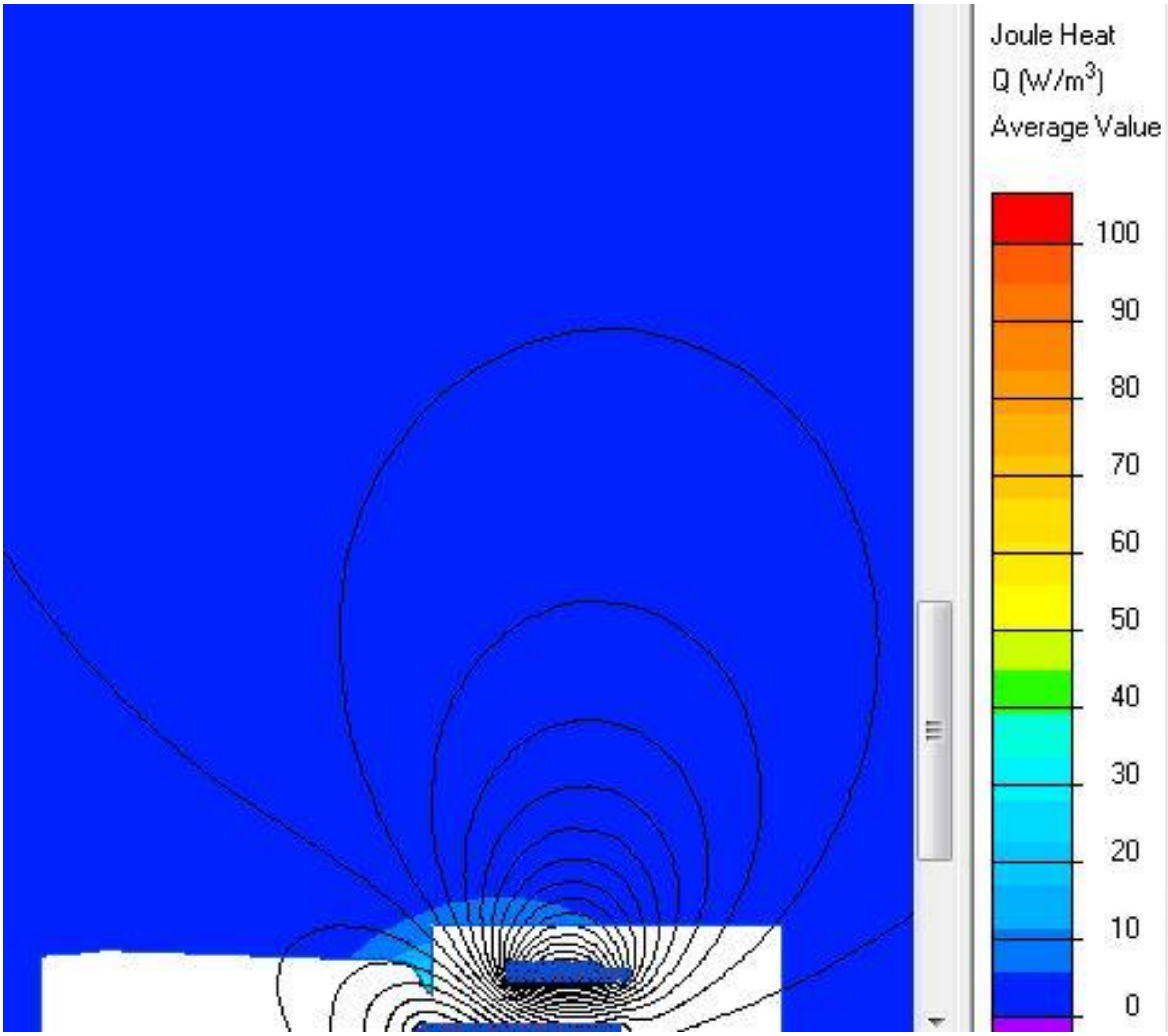

Figure 11
42 KHz 69.4 Ohm load SAR data

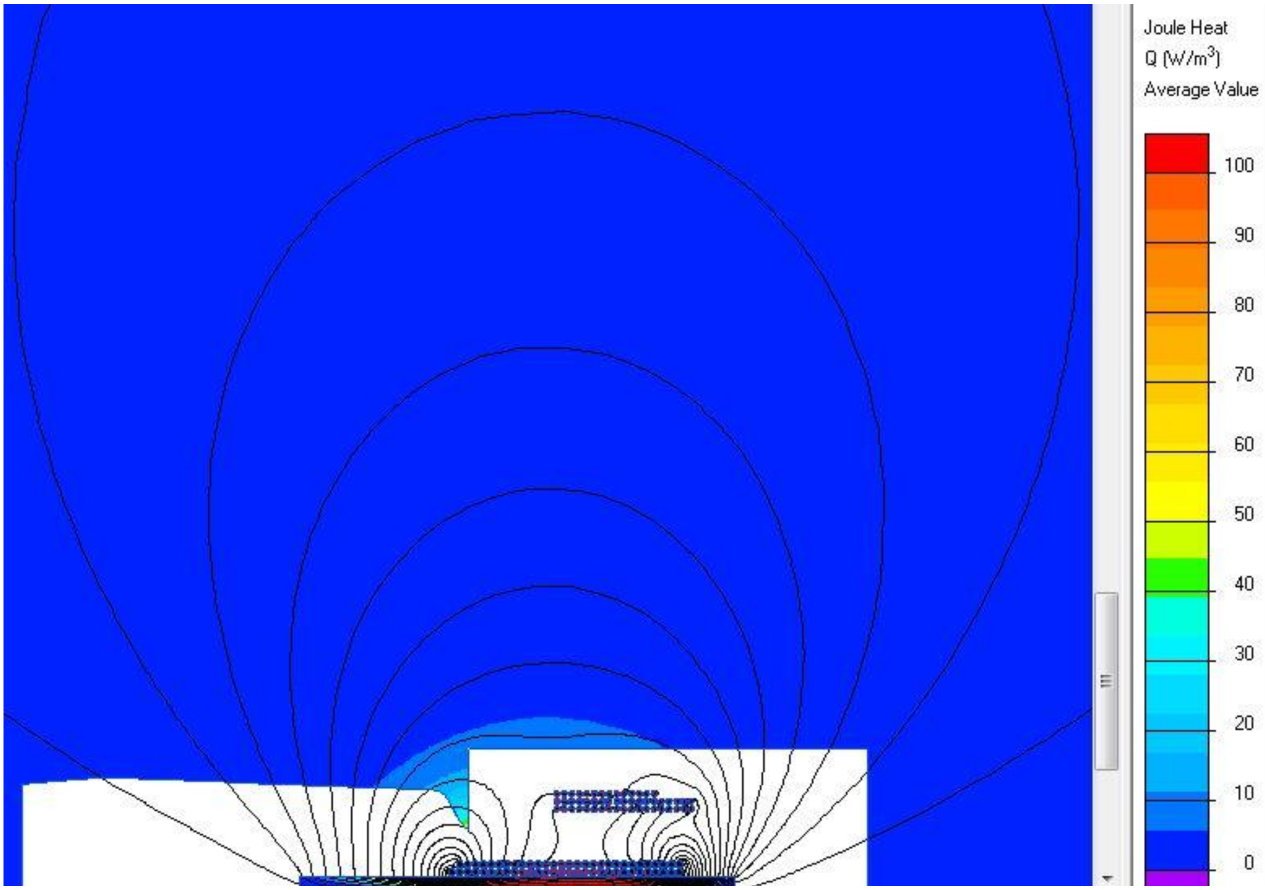

Figure 12
42 KHz 10 kOhm load SAR data

A density of 1g/cm3 was assumed. This implies that the FCC's stated limit of 1.6 W/kg corresponds to 1600 W/m3. The model was configured for 42 kHz operation with the charger plug inserted into the receiver coil / socket. The simulations all predicted the same peak, sub 1g mass, SAR values of 17.5 to 17.6 W/m3 for all loads. This places us well below the SAR limits for 42 kHz operation.

It should also be pointed out that the width of the V shaped notch between the charger plug and the receptacle is <2mm in width. As such no tissue can intrude into this gap.

It should also be pointed out that the geometry of the receiving block is based on the current product. Any future products submitted to a TCB will be designed to have at least these spacings and will refer to this summary document of KDB 880208.

79 kHz SAR data
Unloaded SAR, which could occur in direct contact with tissues, was similarly simulated at 79 kHz. Our simulation was made with elements smaller than 1mm on a side. For this reason, our peak readings should be very conservative. A graphical plot of the power absorbed into tissues with a conductivity of 1 S/m is shown in Figure 13. The peak value on the surface of the

enclosure is 180 W/m3 which is well below the FCC limit. Also, as can be seen in Figure 13, even this exposure is limited to tissues within a few mm of the surface.

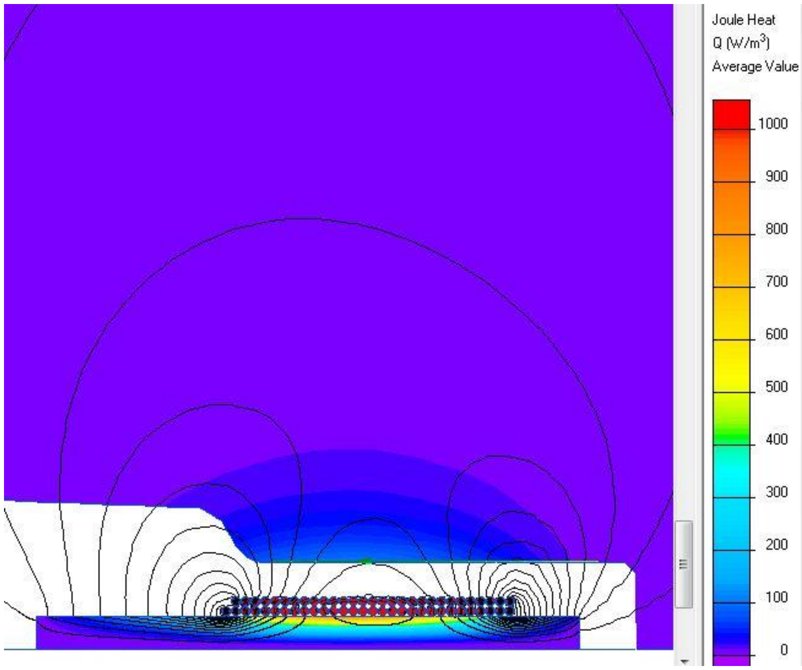

Figure 13
Power absorption with 79 kHz 20.49V drive.

ICNIRP stimulation limit data

As mentioned above, the ICNIRP has recommended limits for E field exposure to prevent the stimulation of nervous tissues. These limits for 42 kHz and 79 kHz are, respectively, 8.2V/m and 15.08 V/m peak. Because induced E fields have been shown to be unaffected by current densities in the physiological range, these simulations were run at 1 S/m so that current density and E field intensity will be the same.

42 kHz E Field Results

Plots of simulated loaded systems at 42 kHz are shown in Figure 14, Figure 15, and Figure 16. Figure 15
Current Density at 42 kHz with 69.4 Ohm load.was inadvertently scaled differently than Figure 14 and Figure 16 so color indications are different.

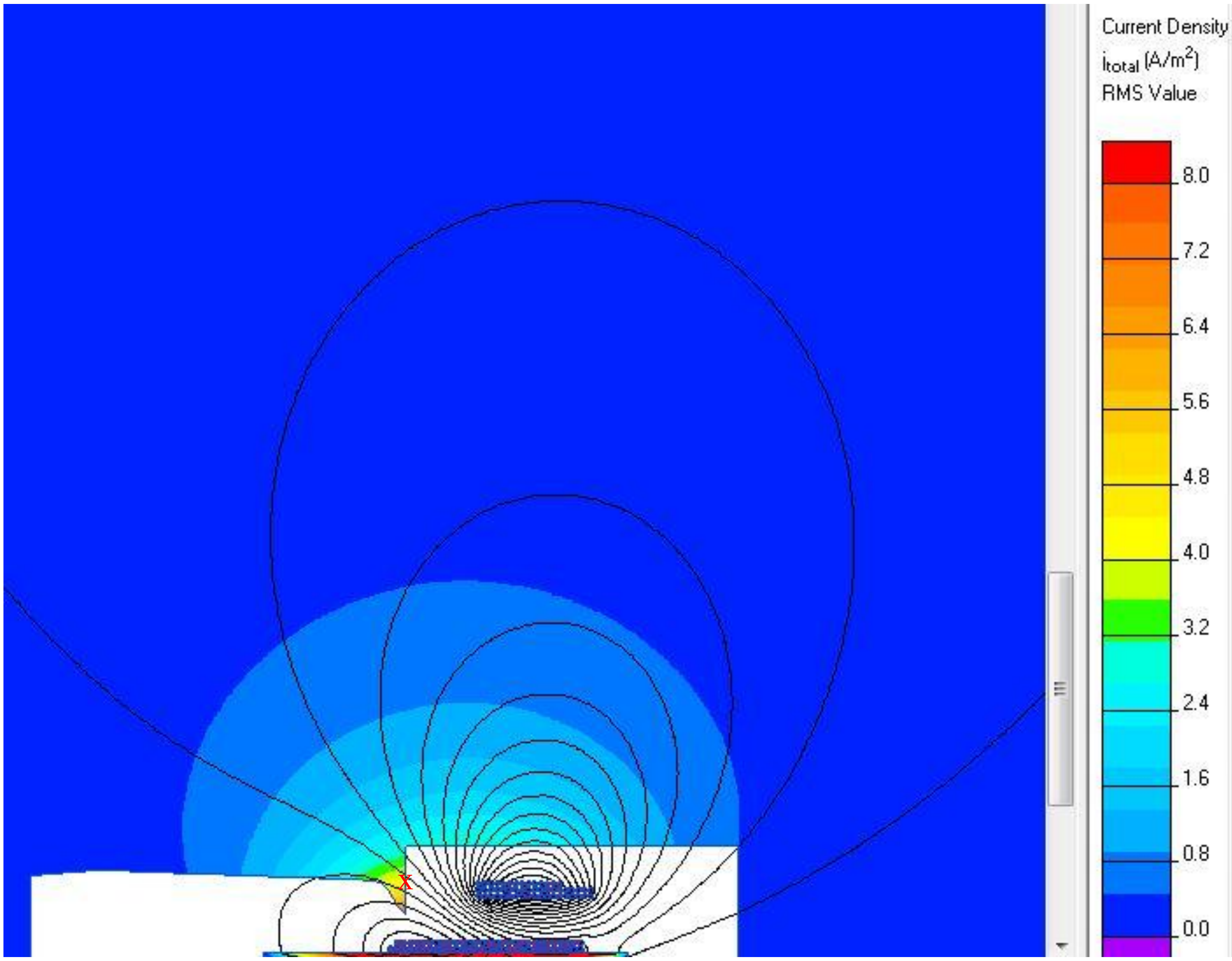

Figure 14
Current Density at 42 kHz with 34.7 Ohm load.

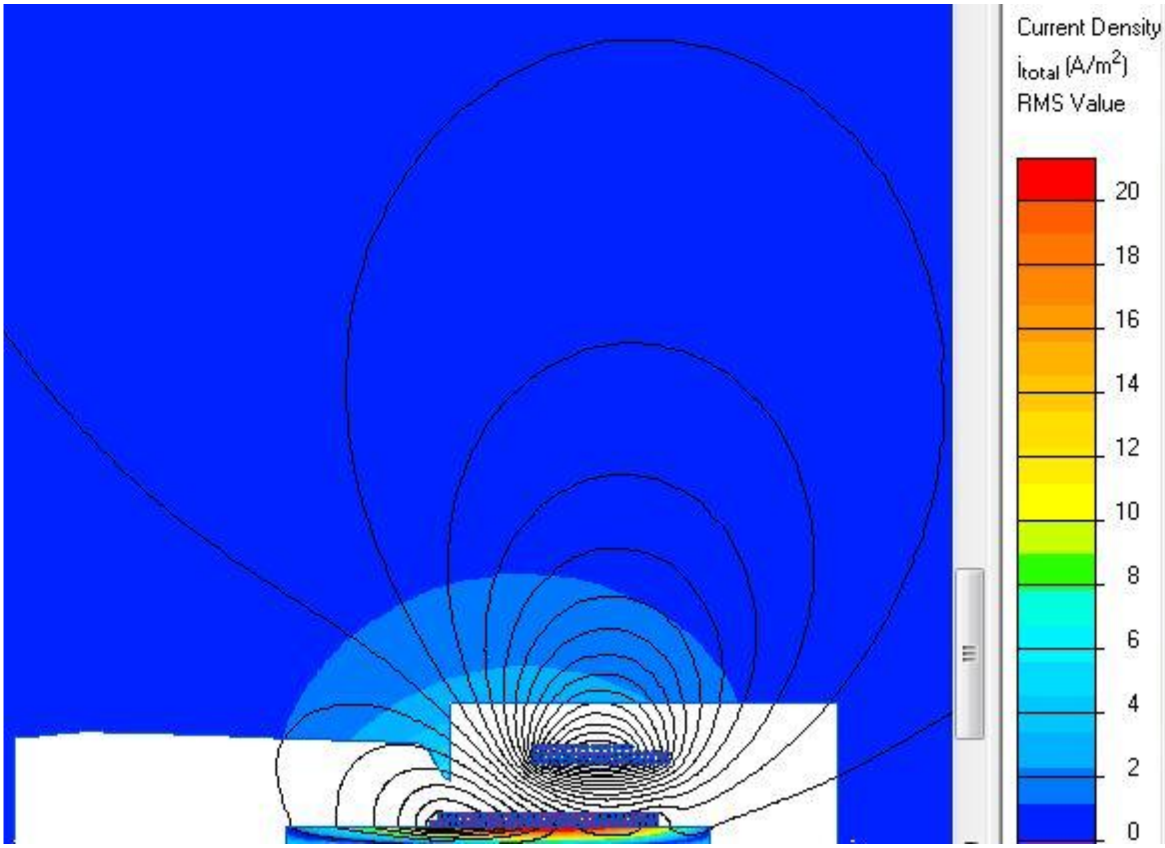

Figure 15
Current Density at 42 kHz with 69.4 Ohm load.

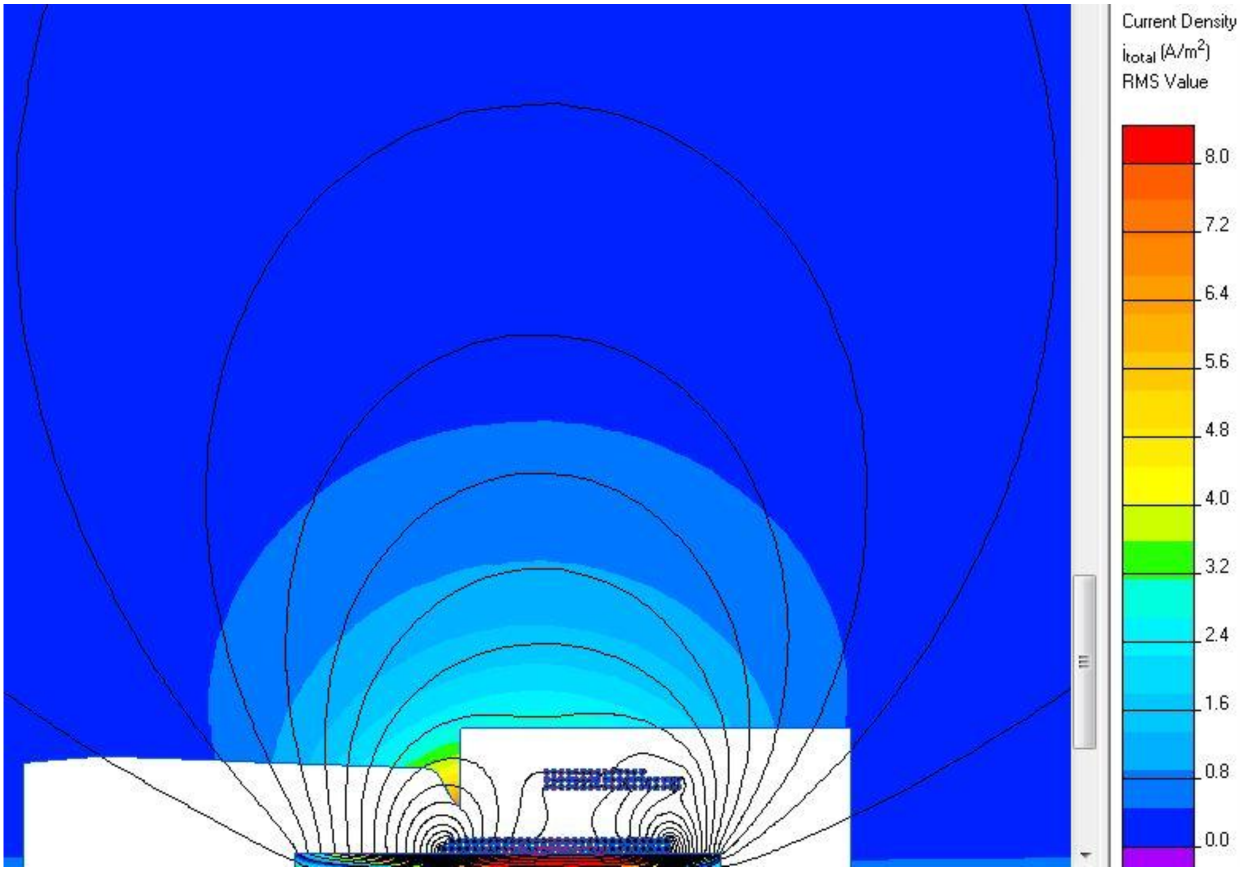

Figure 16
Current Density at 42 kHz with 10 kOhm load.

Maximum peak E fields for each of these loading conditions, as measured at the point indicated by the red x in Figure 14, were 4.2 V/m for all conditions. This is approximately ½ ICNIRP Basic Restriction. Further, the user exposes at most 2 mm of the surface of their finger tip to this maximum value leading to the conclusion that a 10mm cube of tissue (1g at 1g/cc) would have an exposure of well below this peak value. As such, the design poses no risk to users in the plugged in mode with 42 kHz active.

79 kHz E Field Results
A graphical plot of peak E field values at 79 kHz is shown in Figure 17.

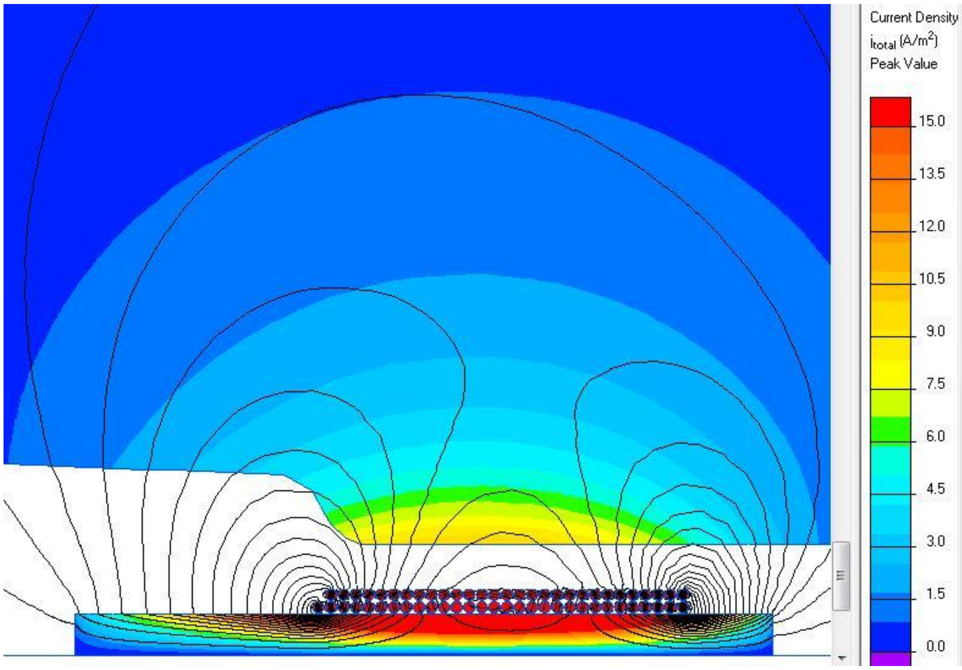

Figure 17
Peak E field with 79 kHz 14.49V drive.

The maximum values seen on the surface of the plug when operating at 79 kHz and 14.49 volts was 9.8V/m. This is well below the ICNIRP limit of 15.08V/m. As with the heating calculations, even this maximum value is limited to within a few mm of the surface of the plug and is a peak value as opposed to an average value. This implies that our design is actually fairly conservative.

Summary
In summary, all power absorption and induced E field intensities were well under the guidelines suggested by the FCC and ICNIRP and generally only extend a few mm from the surface of the charger hardware.

Thank you for your time and assistance,

Sincerely,

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[i] ICNIRP GUIDELINES FOR LIMITING EXPOSURE TO TIME-VARYING ELECTRIC AND MAGNETIC FIELDS (1 HZ – 100 kHZ) PUBLISHED IN: HEALTH PHYSICS 99(6):818-836; 2010

[ii] ICNIRP GUIDELINES FOR LIMITING EXPOSURE TO TIME-VARYING ELECTRIC, MAGNETIC AND ELECTROMAGNETIC FIELDS (UP TO 300 GHZ) PUBLISHED IN: HEALTH PHYSICS 74 (4):494-522; 1998

[iii] ICNIRP STATEMENT GUIDANCE ON DETERMINING COMPLIANCE OF EXPOSURE TO PULSED FIELDS AND COMPLEX NONSINUSOIDAL WAVEFORMS BELOW 100 kHZ WITH ICNIRP GUIDELINES PUBLISHED IN: HEALTH PHYSICS 84(3):383-387; 2003

[iv] The dielectric properties of biological tissues: I. Literature survey, C Gabriel, S Gabriel, et al, Phys. Med. Biol 41 (1996) 2231 – 2249.