A novel device for determining ultrasonic power

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A novel device for determining ultrasonic power

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Abstract. A novel concept for an ultrasonic power meter is presented which utilises the *pyro*-electric effect of a thin membrane of the *piezo*-electric polymer, *pvdf*. One side of the membrane is in intimate contact with a polyurethane-based acoustical absorber. The attenuation coefficient of this material is very high, ensuring that the majority of the ultrasonic energy passing through the *pvdf* membrane is absorbed within a thin layer of the interface, resulting in a rapid increase in temperature. Through the *pyro*-electric effect, this temperature increase results in a voltage across the electrodes of the membrane. Under specific conditions, the generated voltage is proportional to the rate of temperature rise and, immediately after switch on of ultrasound, the rate of temperature rise is proportional to the delivered ultrasonic power. This paper describes details of the concept, and includes theoretical calculations of the expected behaviour. Proof-of-concept is demonstrated through careful studies of several low-megahertz NPL reference therapy-level transducers, covering an applied power range of 250 mW to 8 W. Results are encouraging, suggesting that this novel solid-state power meter concept holds considerable promise for a rapid, simple, relatively low cost, power measurement method, appropriate for use at the physiotherapist level.

1. Introduction

Physiotherapy ultrasound is widely used to treat soft tissue injuries, with over a million treatments being administered within the United Kingdom alone. Despite its wide-spread use, and the potential for harmful thermal affects due to the high applied ultrasonic powers (up to 12 W), the calibration status of this type of equipment has long been very poor [1]. Perhaps the most important reason for this is a complete lack of traceable measurement methods which can be applied quickly, and cost-effectively, at the physiotherapy level i.e. at the point of delivery of the treatment. The standardized method for determining ultrasonic power is through the measurement of radiation force [2], and high quality radiation force balances are commercially available which permit power to be determined with an uncertainty typically better than ±10%. However, these devices tend to be fairly expensive requiring a reasonable level of expertise to set-up and operate, characteristics which render them inappropriate for the end user of the equipment. There is therefore a need for a new type of measurement device which possesses a number of attributes. It should be:

- compact and simple in construction;
- easy and quick to use;
- low-cost
- ideally, it should provide an output related to an ***absolute*** quantity such as the ultrasonic power.

This paper introduces a novel, *pyro*-electric based measurement concept, which potentially meets many of these requirements. The device described has been the subject of a UK Priority Patent Application Number 0127529.6.

2. Outline of new concept

Figure 1 represents a simple schematic of the new sensor. It comprises two components: a thin layer of *pyro*-electric material made from a membrane of poly-vinylidene fluoride (*pvdf*). This is backed with a layer of material that it is extremely attenuating to ultrasound at megahertz frequencies, transducer and membrane being coupled through a short water path. When the transducer is switched ON, acoustic energy passes through the membrane and is absorbed within a very short distance of the membrane-absorber interface, leading to a rapid local heating of the material close to the membrane. The *pyro*-electric nature of the membrane results in a voltage being generated across its electrodes and the operation of the device centres on the measurement of this *pyro*-electric voltage.

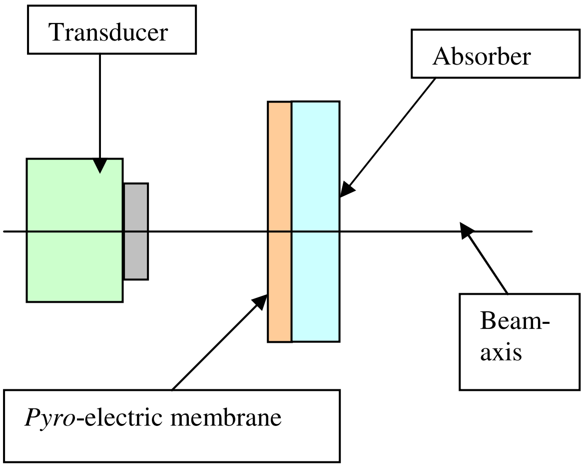

Figure 1: Schematic of the make-up of the new sensor, showing the thin *pyro*-electric membrane made from *pvdf*, backed with an acoustically absorbing material.

3. Theoretical description

Taking the experimental configuration described in Figure 1 and considering the temperature rise in the absorber, then, by superimposing heat sources over a given volume, the temperature increase at time *t*, $\Delta T(t)$, may be represented by [3]:-

$$\Delta T(t) = \frac{1}{2K} \int_0^{\varepsilon} \int_0^{\frac{d}{2}} \frac{q_V(z)}{r(z)} \operatorname{erfc}\left(\frac{r}{\sqrt{4\kappa t}} \right) x dx dz \tag{1}$$

where: $\Delta T(t)$ represents the temperature elevation above ambient level; q_V is the rate of heat production rate per unit volume; κ is the thermal diffusivity; K is the thermal conductivity and ε is the thickness of the heat source. In Equation 1, d is the diameter of the heat source and r is the distance from the source; $r = \sqrt{(\varepsilon - z)^2 + x^2}$. If we consider only primary *pyro*-electric effects, then the sensor output charge $q(t)$ is directly proportional to the temperature rise [4] such that:

$$q(t) = pA\Delta T(t) \tag{2}$$

where p is the pyroelectric charge coefficient and A is the sensor surface area. When the sensor is connected to a voltage amplifier of input resistance R_a, the equivalent electrical circuit may be described in terms of the following differential equation [4]:

$$\frac{pA}{C}\frac{d\Delta T(t)}{dt} = \frac{V}{RC} + \frac{dv}{dt} \tag{3}$$

where R is the parallel combination of R_p (the internal electrical resistance of the sensor) and R_a; C is the capacitance of the layer of *pvdf*; V is the voltage at the input stage of the amplifier. The important feature of Equation 3 is that, when the product RC is sufficiently small, the voltage V is proportional to the rate of change of temperature. This condition can be achieved by choosing an amplifier with suitable input impedance. At times immediately following switch ON of the transducer, the generated *pyro*-electric voltage is proportional to the rate of change of temperature rise, which is itself proportional to delivered acoustic power, as the response is integrated over a surface which is substantially bigger than the ultrasonic beam.

Figure 2 illustrates an example calculation carried out at a frequency of 1 MHz, and for powers of 1 W and 5 W. The transducer has been 'activated' at approximately 0.9 seconds, and the *pyro*-electric voltage increases rapidly to reach a maximum after approximately 200 ms. The maximum excursion upon switch ON is proportional to the generated ultrasonic power. Beyond this, heat generated within the medium at the interface starts to conduct away leading to a reduction in the rate of temperature rise, and a consequent reduction in the magnitude of the *pyro*-electric voltage until t = 2.9 seconds, when the ultrasound source is switched OFF. Upon switch OFF, the material starts to cool, and the *pyro*-electric voltage eventually changes sign. It should be noted that one could equally take the change in voltage at switch OFF, as a measure of the delivered acoustic power.

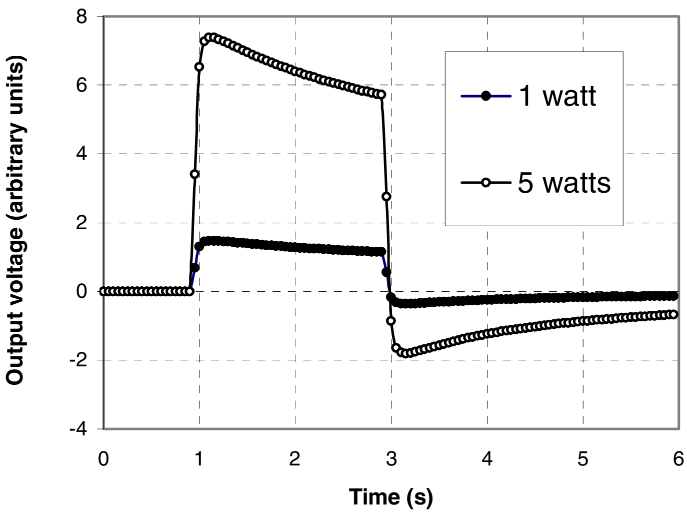

Figure 2: Results of calculations of the response of the new sensor concept to ultrasound at a frequency of 1 MHz, at two applied powers

4. Materials and methods.

It is only possible to present a few key results whose aim is to demonstrate proof-of-principle for the sensor. A few points to note regarding the measurements described in Section 4 are:

• the power meter concept is currently being commercially exploited by Precision Acoustics, and measurements have been carried out on the 'front-end' (sensor) of one of their devices. This uses an angled *pvdf* membrane (20°) designed to prevent reflections returning coherently back to the front face of the treatment head. The well containing the membrane and water couplant is 60 mm in diameter and of mean depth 24 mm;

• the backing material is based on the NPL absorber described in Zeqiri and Bickley [5];

• the *pvdf* membrane was 52 µm thick;

• well characterized powers have been applied using NPL reference therapy-level transducers, spanning the frequency range 1 to 3 MHz;

• generated *pyro*-electric voltages were amplified using a Bruël and Kjær Type 2651 Charge amplifier (1 mV pC^{-1} setting) and recorded using a Tektronix TDS 7104 oscilloscope.

5. Results

For tests, the applied power was determined using the NPL reference therapy-level balance. The transducer was then transferred to the device, and positioned centrally over its membrane with the transducer body being nominally vertically aligned. Measurements on the new sensor were undertaken using a well-defined exposure protocol which involved leaving the power to the treatment switched ON for only two seconds, and recording the *pyro*-electric voltage 2 seconds prior to switch ON and 6 seconds subsequent to switch OFF. Following these tests, the delivered power was re-checked by returning the transducer to the NPL therapy-level radiation force balance.

It should be noted that the *pyro*-electric waveforms shown in the subsequent figures are 'single-shot'. However, 'shot-to-shot' repeatability of the peak measurements is excellent, being typically a few per cent. Figure 3 shows the results of two measurements made using a small diameter 3 MHz transducer (effective radiating area 37 mm^2), operating at two powers: 0.537 W (Low) and 3.5 W (High). As is evident from Figure 3, the time variation of the *pyro*-electric waveform for the two measurements is qualitatively very similar to the predicted dependence shown in Figure 2, a well-defined peak being prominent a few hundred milliseconds after switch ON. In order to verify that the general shape of the observed waveform did not change for the two powers, these have been normalized to the peak voltage derived using the high power setting. In fact, the two observed peak-voltages generated were 0.38 V (Low), and 2.53 V (High) respectively, giving a ratio of 6.66. In comparison, the ratio of the applied powers was 6.52, an agreement of approximately 2%. Confirmation of this excellent linearity, has been obtained from through extensive measurements covering a power range of 250 mW to 8 W.

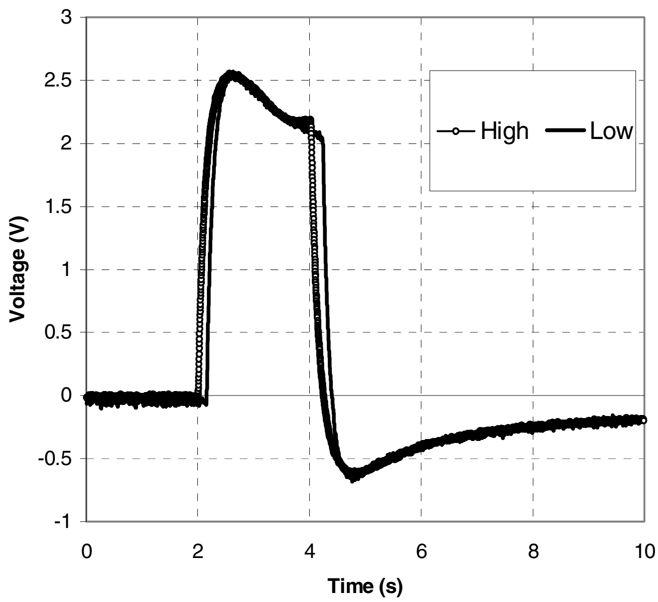

Figure 3: Comparison of the *pyro*-electric waveform generated by the new sensor in response to the ultrasonic field of a 3 MHz therapy transducer operating at two power levels: 0.537 W (Low) and 3.5 W (High). The values derived from the lower power have been scaled to the peak value derived for the higher setting

Figure 4 shows the variation of the *pyro*-electric voltage for both the small 3 MHz transducer, and a larger diameter reference transducer operating at the same frequency, whose effective radiating area is 346 mm^2. Each of the transducers was driven to deliver 3.5 W (±2%) with the responses of the *pyro*-electric sensor compared. The two waveforms are shown in Figure 4, where the peak-voltages agree to within ±2%. Additionally, the two traces are very similar in shape. This is a very important finding, as it indicates that the very different transducer beam characteristics appear to have little influence on the *pyro*-electric voltage which is primarily governed by the applied power.

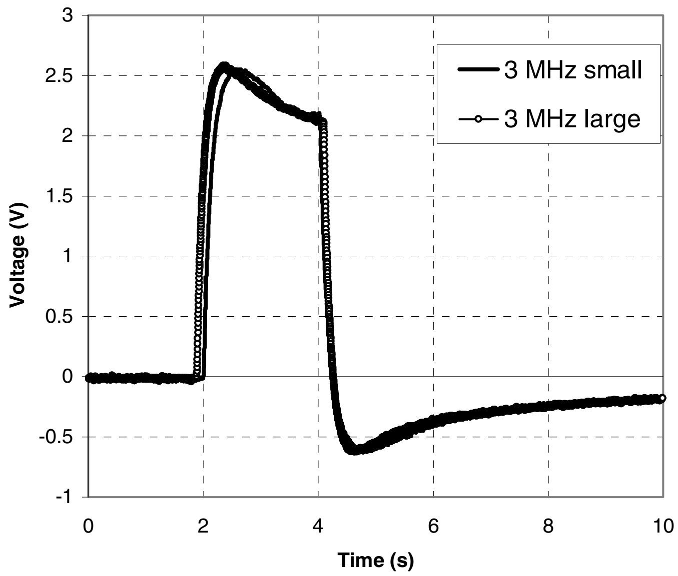

Figure 4: Comparison of the *pyro*-electric waveform generated by two transducers operating at 3 MHz and output power 3.5 W. The transducers have different effective radiating areas: 37 mm^2 (Small) and 346 mm^2 (Large)

6. Discussion

The results shown in Section 4 illustrate proof-of-principle for the new method, demonstrating the linearity of the device, and also that the output of the novel sensor appears to depend only relatively weakly on the beam-shape, for times immediately after switch ON of the electrical excitation to the transducer. However, these tests have been carried out at 3 MHz, where the absorption coefficient of the absorber is in excess of 90 dB cm^{-1}. This means that approaching 88% of the acoustic energy will be absorbed within 1 mm of the front surface of the acoustical absorber

Performance tests have also been carried at a frequency of 1 MHz using NPL reference transducers, and in this case a well-defined maximum characterising the voltage responses shown in Figures 3 and 4 are not produced. Rather, the *pyro*-electric voltage increases initially rapidly up to 500 ms after switch ON, and thereafter continues to increase more slowly by a further 5% to 10% over the final 1.5 seconds of ON time. This behaviour may be understood by considering the attenuation coefficient of the polyurethane based material at 1 MHz, which is only 30 dB cm^{-1}. In this case, the top 1 mm of the polyurethane absorbs only 50% of the power generated, resulting in a greater acoustic power being deposited deeper within the absorber, creating thermal sources which heat the membrane at significant times after switch ON and, indeed, even after switch OFF. This is evidence that the acoustic attenuation at 1 MHz is not sufficiently high. For future developments, it will be relatively straightforward to further develop the polyurethane absorber in order to boost this attenuation so that it is greater than 70 dB cm^{-1} at 1 MHz.

6. Conclusions

This paper has described a novel concept for a solid-state ultrasonic power meter, which utilises the *pyro*-electric effect of a thin membrane of polymeric material (*pvdf*). The membrane is backed by an acoustical absorber which absorbs most of the acoustic energy incident upon it, leading to a rapid increase in the temperature at the interface between the two materials. Proof-of-concept measurements have been presented which were undertaken at 3 MHz. These illustrate a response which is linear with applied power, and is relatively insensitive to the effect of transducer diameter (beam-shape). However, studies also indicate that the intrinsic attenuation in the material backing the *pyro*-electric membrane needs to be increased for frequencies around 1 MHz. Studies of the performance of this novel sensor will continue.

7. References

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