Novel Passive Implantable Atrial Defibrillator Using Transcutaneous Radiofrequency Energy Transmission Successfully Cardioverts Atrial Fibrillation

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Background—Conventional methods for cardioversion of atrial fibrillation (AF) to sinus rhythm have numerous difficulties. A novel method for cardioversion using the passive implantable atrial defibrillator (PIAD) was tested in acute animal models. This device does not have a battery or a capacitor to store energy and is activated by transferring RF energy across the skin from an external transmitter to the subcutaneously implanted defibrillator. On activation, a novel monophasic shock waveform with 5% tilt is delivered to the heart via 2 intracardiac defibrillation leads.

Methods and Results—Cardioversion attempts with the device were assessed in 2 phases: a feasibility and efficacy study and randomized comparison against standard waveforms. Defibrillation leads were placed transvenously into the distal coronary sinus and the right atrial appendage. These were connected to the subcutaneously implanted PIAD. Sustained AF was induced by rapid atrial pacing. The transmitter coil was placed on the skin overlying the defibrillator, and defibrillation synchronized to the R wave was attempted. The method was found to be efficacious at very low voltage and energy, with 100% cardioversion success observed for 10-ms 100-V shocks (mean energy, 1.54 ± 0.02 J). The PIAD waveform had a higher cardioversion success rate than a truncated, 70% tilt monophasic exponential pulse (100 V, 100% versus 78.0 ± 7.57%; P = 0.001). There were no postshock complications.

Conclusions—Considering these animal results, this method is promising for cardioverting AF in symptomatic patients. 

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Key Words: arrhythmia ■ fibrillation ■ cardioversion

Atrial fibrillation (AF), the most common sustained cardiac arrhythmia, continues to be a challenging rhythm to treat in clinical practice. It is generally accepted that conversion to and maintenance of sinus rhythm is the preferred management option. We report a novel way of defibrillating the heart that may have wider applications in clinical medicine.

Cardioversion can be performed pharmacologically or by DC shock applied transthoracically and, more recently, transvenously to the heart. The transthoracic method is efficacious but requires high energies and anesthesia and may be detrimental to the heart.1–6

In both animal and clinical studies, the transvenous route for cardioversion has proved to be safe and efficacious at significantly lower energy levels while requiring minimal or no sedation.11,14

This led to the development of the automatic implantable atrial defibrillator.15,16 This subcutaneously implanted device is battery powered and delivers defibrillation shocks from capacitor-stored energy to the atria through 2 transvenously placed defibrillation leads. It is indicated primarily for patients who have recurrent, poorly tolerated AF that is not controlled pharmacologically. Although it is efficacious at cardioverting AF, the primary difficulties with this automatic device include poor tolerance in some patients to shock delivery,14 overall cost of the device, the need for backup ventricular pacing to counter potential shock-induced brady-arrhythmias, and dependence on complex software and algorithms for AF and ventricular fibrillation detection and R-wave–synchronized shock delivery.

We therefore developed a novel method for cardioversion of AF: the passive implantable atrial defibrillator (PIAD). This animal study assessed the feasibility, efficacy, and safety of this method for cardioversion. We further hypothesized and tested that the novel, low-tilt (5%) monophasic waveform delivered by the PIAD will be at least as efficacious as monophasic 70% tilt pulses.

Passive Implantable Atrial Defibrillator

This unique implanted defibrillator (Figure 1) does not have a battery or a discharging capacitor and is powered transcutaneously by inductive coupling using RF energy.
Figure 1. PIAD (55 mm maximum diameter x 70 mm length x 8 mm thick). Output lead connects to defibrillation leads.

Transfer of energy through the skin uses an external coil system to transmit and a stand-alone (passive) internalized coil system to receive the energy (magnetic pulse field) via an inductive link. The PIAD uses an “on-off” pulsed RF power source operating at 7.2 MHz and connected to an RF transformer. The RF transformer refers to a series-tuned primary placed on the body surface (transmitter) and a parallel-tuned secondary (implanted subcutaneously) acting as the “receiver.” On activation, the implanted defibrillator delivers a novel monophasic DC shock waveform with low tilt (5%) and rounded edges (Figure 2).

A radio transceiver (Yaesu SSB Transceiver FT-101E, Yaesu Musen Co Ltd) generated the RF energy (Figure 3). The RF tuner (MFJ-989c Tuner, MFJ Enterprise Inc) optimizes coupling between the hand-held transmitter and the receiver units. An RF “keying” unit was connected to the RF generator, which enables pulsed (on-off) generation of the RF signal and setting of the pulse width of the shock waveform. A custom-built R-R interval detector and R-wave synchronization device were set to deliver an R-wave–synchronized shock.

Methods
All work was carried out in accordance with the United Kingdom Animals (Scientific Procedures) Act of 1986.

The experiments were carried out in 2 phases. In phase 1, we investigated the device in an experimental sheep model of AF (n=10; mean weight, 58.0 ± 7.1 kg) using varying voltage and shock waveform pulse-width settings to evaluate the feasibility, efficacy, optimum shock pulse width, and safety of the device at cardioversion of AF. In phase 2, the efficacy of the novel PIAD waveform was randomly compared against 2 monophasic waveforms in an experimental porcine model of AF (n=10; mean weight, 53.8 ± 0.3 kg).

Equipment
In phase 2, a Ventritex HVSO2 defibrillator (160 μF) was used to deliver 2 monophasic waveforms: a truncated exponential pulse with 70% tilt and a rounded-edge truncated exponential pulse with 60% tilt (Figure 2). The latter had been reported to produce lower defibrillation energy thresholds than the standard truncated exponential waveform.

Probes connected to the interface device captured the delivered voltage and current (Figure 3). These were then plotted as voltage–time and current–time curves on a digital oscilloscope (TDS 420A, Tektronix Inc) and stored for further analysis of the delivered energy and impedance.

Procedure
The animals were anesthetized (titrated to response) by pentobarbital (30 mg/kg IV; Sagatal, Rhône Mérieux Ltd) and ventilated with room air (model 16/24, the “Ideal” respirator, CF Palmer [London] Ltd).

Defibrillation leads were inserted via the internal jugular vein, one positioned against the lateral wall of the right atrium and the other in the distal coronary sinus, under fluoroscopic guidance (Figure 4). The internal impedance between the leads (measured) and lead positioning were checked frequently throughout the experiment to ensure stability. Intravenous heparin (100 U/kg) was used for anticoagulation.

Figure 2. Standard monophasic (SM), rounded monophasic (RM), and PIAD waveforms are shown. Compared with SM and RM, PIAD waveform has negligible tilt (5%), with rounding of both leading and trailing edges, and has a slow-rising (R) and -falling (F) pulse. Inset, Current waveform (CW) for PIAD. T indicates oscilloscope trigger.
The PIAD was then implanted subcutaneously in the left lower parasternal area and connected to the defibrillation leads via an interface device (Figure 3). Burst right atrial pacing (100 Hz, 2-ms duration, 5 V for 5 seconds) induced AF via a right atrial bipolar catheter and a Grass stimulator (Grass S44 Stimulator). After induction and confirmation of AF, the RF transmitter was placed on the skin overlying the PIAD, and cardioversion was attempted.

The surface ECG (lead II), synchronization spikes (timed with the R wave), and arterial blood pressure tracings were displayed on a polygraph (Gould 2400S). Body temperature, oxygenation, and serum electrolytes were monitored and maintained within normal limits.

The mean peak voltage, peak current, energy, and impedance were calculated from the stored data. The energy delivered was calculated by summing the areas under the instantaneous voltage–current product curves at 0.1-ms intervals. The ECG was assessed for evidence of proarrhythmia or atrioventricular block after defibrillation.

**Protocol**

**Phase 1**: We randomly assessed the PIAD at 3 voltages of 50, 75, and 100 V. The 50- and 100-V settings were assessed in 10 sheep at pulse widths of 5, 6, 8, 10, 12, 15, 20, and 30 ms. The 75-V setting was assessed in 8 of the 10 sheep and at the 8-, 10-, 12-, 15-, and 20-ms pulse widths.

**Phase 2**: The PIAD, the standard monophasic, and the rounded monophasic waveforms were compared randomly at 50 and 100 V (pulse width, 10 ms) in 10 pigs.

For both phases, 5 attempts at cardioversion were made at each voltage and pulse-width setting to assess the percentage success of the shock. If an attempt was successful, AF was reinduced.

**Definitions**

Sustained AF (>30 seconds in duration) was defined as a narrow QRS complex rhythm without P or flutter waves and with an irregularly irregular ventricular response. The shock results were defined as follows: success = reversion to sinus rhythm within 3 seconds of shock delivery; failure = persistence of AF or conversion to another arrhythmia.

**Statistical Analysis**

All data are expressed as mean ± SEM. The nonparametric Friedman test was used to evaluate differences among the success rates, mean energy, peak voltage, and peak current within waveforms and set voltages. If a significant result was seen, the nonparametric 2-related Wilcoxon signed rank test was used to evaluate comparisons between 2 waveform and voltage settings. All statistical analyses used SPSS statistical software (SPSS version 10). Differences were considered significant at a value of $P<0.05$.

**Results**

**Phase 1**

A total of 1000 shocks was delivered, all synchronized to the R wave. There were no significant differences in the direct internal impedance measurements between the leads (mean impedance, 51.8±1.1 Ω).

**Success Rates**

We observed a mean success percentage of 100% for the 10-ms 100-V setting, with a mean energy delivered of 1.54±0.02 J (Figure 5). This was significantly more efficacious than the 8-ms 100-V setting (88±5.3%; 1.22±0.01 J; $P=0.014$), the 10-ms 75-V setting (72.5±10.7%; 0.89±0.01 J; $P=0.001$), and the 15-ms 75-V setting (82.5±10.3%; 1.38±0.01 J; $P=0.008$). Although the 10-ms 100-V setting had a better success rate than the 20-ms 75-V setting (97.5±2.5%; 1.87±0.02 J), this did not reach statistical significance. The high success rates were also maintained for longer pulse widths (12, 15, 20, and 30 ms) at the 100-V setting compared with the 10-ms setting, but there was a progressive increase in delivered energy.

For the 75-V setting, there were gradual improvements in success rate with increasing pulse widths, with 97.5±2.5% (1.87±0.02 J) achieved for the 20-ms ($P=0.001$) setting. The 75-V setting was also more efficacious than the 50-V at the 12-ms (82.5±7% versus 56±8.8%; $P=0.025$) and 20-ms (97.5±2.5% versus 74±9.9%; $P=0.005$) pulse width settings. Although the 75-V setting was more efficacious than the 50-V at the 8-, 10-, and 15-ms pulse width settings, statistical significance was not achieved.

There were gradual improvements in efficacy for the 50-V setting with increasing pulse widths ($P<0.0001$). A plateau was reached at the 20-ms setting, with 74±9.9% (0.87±0.01 J) success observed. Although there were gradual improvements in success rate from the 12-ms to the 30-ms pulse width setting, significant differences were not achieved.

**Shock Parameters**

The peak voltage for the 100-V 10-ms setting, at which the optimum success rate was observed, ranged between 98.0 and
116.0 V (mean, 104.5±0.3 V) with current ranging between 2.0 and 2.5 A (mean, 2.3±0.0 A) and calculated impedance ranging from 41.8 to 55.6 Ω (mean, 46.2±0.2 Ω) (Table). With the measured mean impedance remaining relatively unchanged throughout each experiment (51.8±1.1 Ω), a corresponding increase in delivered current was observed with increasing delivered voltage.

**Phase 2**

A total of 300 shocks were delivered, with no difference in the internal measured impedance between the leads (48.10±0.95 Ω).

**Delivered Peak Voltage, Current, and Impedance for Phases 1 and 2**

<table>
<thead>
<tr>
<th>Voltage Setting, V</th>
<th>50</th>
<th>75</th>
<th>100</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Phase 1</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Voltage, V</td>
<td>56.1±0.2</td>
<td>80.4±0.3</td>
<td>104.5±0.3</td>
</tr>
<tr>
<td>Current, A</td>
<td>1.1±0.0</td>
<td>1.7±0.0</td>
<td>2.3±0.0</td>
</tr>
<tr>
<td>Impedance, Ω*</td>
<td>49.3±0.2</td>
<td>48.0±0.3</td>
<td>46.2±0.2</td>
</tr>
<tr>
<td><strong>Phase 2</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Voltage, V</td>
<td>53.3±0.5</td>
<td>100.4±0.6</td>
<td></td>
</tr>
<tr>
<td>Current, A</td>
<td>1.2±0.0</td>
<td>2.3±0.0</td>
<td></td>
</tr>
<tr>
<td>Impedance, Ω*</td>
<td>43.4±0.5</td>
<td>43.8±0.5</td>
<td></td>
</tr>
</tbody>
</table>

**SM** indicates standard monophasic waveform; **RM** rounded monophasic waveform; and **PIAD** PIAD waveform. Data are mean±SEM.

*Peak impedance calculated from the measured peak voltage and current by use of the following formula: voltage (V) = current (A) × impedance (Ω).

**Success Rates**

At the 100-V 10-ms setting, the PIAD was significantly more efficacious (100%; 1.74±0.02 J) than the 100-V standard (78.0±7.57%; 0.77±0.01 J; P=0.001) or the 100-V rounded monophasic (92.0±4.42%; 1.38±0.01 J; P=0.046) waveform at the same pulse widths (Figure 6). The PIAD 100-V setting was also more efficacious than any of the 50-V waveform settings (P<0.003).

At the 50-V 10-ms setting, rounding of the standard waveform (rounded monophasic) improved the success rates significantly compared with the standard monophasic waveform (46±10.8% versus 30.0±10.4%; P=0.02). Although the mean success rate for the 100-V rounded waveform setting was better than the standard (92.0±4.4% versus 78±7.6%), it did not reach statistical significance.

There was no statistical difference in percentage success when the PIAD 50-V setting (80±8.4%; 0.48±0.02 J) was compared with the 100-V standard or the 100-V rounded monophasic waveform settings. These were observed, despite the PIAD 50-V setting delivering a significantly lower mean energy (100-V standard, 0.77±0.01 J; 100-V rounded, 1.38±0.01 J; P<0.0001).
Cardioversion Parameters
The delivered peak voltage for the 100-V PIAD setting ranged from 89 to 106 V, with a mean of 99.7 ± 0.7 V (Table, phase 2). There was no significant difference between the 100-V PIAD setting and the standard monophasic waveform (range, 92 to 105 V; mean, 100.4 ± 0.6 V), but a slightly lower delivered voltage was observed compared with the 100-V rounded waveform (range, 99 to 109 V; mean, 105.3 ± 0.3 V; P < 0.0001). The mean delivered voltage of the rounded waveform was also higher than the standard at the 100-V setting (P < 0.0001).

Safety
No arrhythmic or hemodynamic complications occurred in either phase of the study.

Discussion
Efficacy of the PIAD
The optimum setting for successful cardioversion of AF using the PIAD is at the 10-ms 100-V setting (100% success; 1.54 ± 0.02 J). However, a good success rate was achieved both at the 50-V 10-ms setting (50 ± 9.1%; 0.43 ± 0.01 J) and at the 75-V 10-ms setting (72.5 ± 10.7%; 0.89 ± 0.01 J). Unlike in ventricular defibrillation, 100% success is not of paramount importance in atrial defibrillation; thus, good success rates can be achieved at low voltages and energies.

The PIAD 100-V setting is also more efficacious than the standard or the rounded monophasic waveforms. Furthermore, there was no statistical difference in percentage success when the PIAD 50-V was compared with the 100-V standard or rounded monophasic waveforms, despite delivering a significantly lower mean energy (Figure 6).

The gradual improvement in success rates observed with increasing pulse width with the PIAD system contrasts with the findings of other investigators using a capacitor discharge transvenous defibrillator. Cooper et al. suggested that shock durations > 10 ms have, in patients, an adverse effect on the defibrillation efficacy of monophasic waveforms (requiring higher voltage and energy). The PIAD waveform, unlike a capacitor discharge truncated exponential monophasic waveform, has negligible tilt (plateau phase of waveform) and thus does not have a decaying tail (Figure 2). The decaying tail is suggested to be profibrillatory. Therefore, for a given duration of the PIAD waveform, all of the pulse width remains above the threshold voltage (minimum peak voltage to achieve successful cardioversion), maintaining a good success rate.

Safety of the PIAD
No arrhythmic or hemodynamic complications occurred for any of the shocks delivered.

Meticulous care was taken to ensure appropriate sensing of and triggering to the R waves before shock delivery in preexperiment dummy load defibrillations. Safe triggering (set to trigger on an R wave with the preceding R-R interval of ≥400 ms) reduces the risk of shock-induced arrhythmia secondary to short R-R intervals, and shock triggering will require meticulous attention when the PIAD is further developed for human application.

Postshock bradycardia or atrioventricular block, often seen in high-energy transthoracic cardioversions, can occur with transvenous defibrillation. In our study, the absence of postshock complications, with good success rates, may be a result of the low currents, voltages, and energies delivered, with the defibrillation leads checked frequently for stability. Poor lead positioning has been associated with an increased risk of complications.

Transcutaneous RF Energy Transfer
Schuder et al. in 1971, first described high-energy transcutaneous RF energy transfer (1 kW) using continuous RF power at 428 kHz to an implanted coil device (11 cm diameter) and a similarly sized transmitter coil (water-cooled) in a dog model. The authors proposed that the resultant energy might be used to recharge batteries for implanted artificial hearts. This mode of energy transfer (continuous), however, would be neither appropriate nor necessary for transvenous cardioversion of cardiac arrhythmias, delivering pulses of 10-ms duration and ~200 W power.

The PIAD is powered by a 7.2-MHz pulsed RF source, with an on-off pulsed RF power connected to a RF transformer. This frequency range enabled the development of a smaller, patient-compliant, implant coil (transmitter diameter, 5 cm; implant diameter, 3 cm). Unlike the findings of Schuder et al, heat production is negligible because of the pulsed nature of the energy delivery (each pulse equates to 1 delivered shock).

Voltage and Current Versus Energy
In this study, we observed that despite a significant increase in the mean energy delivered for the 30-ms 50-V setting compared with the 8-ms 75-V setting (1.28 ± 0.02 versus 0.65 ± 0.01 J; P < 0.0001), the percentage success rates were similar (74.0 ± 9.5% versus 70.0 ± 16.9%). A similar finding was observed when the 30-ms 50-V setting was compared with the 20-ms 50-V setting (energy, 1.28 ± 0.02 versus 0.87 ± 0.01 J; P < 0.0001; success, 74.0 ± 9.5% versus 74 ± 9.9%).

Therefore, it may well be that the voltage and current amplitude and waveform determine the outcome of cardioversion rather than the mean energy delivered to the fibrillating heart.

Waveform Tilt
Waveform tilt describes the rate of decay of the shock voltage from its peak (leading edge) to the voltage at truncation (trailing edge), usually expressed as a percentage [peak voltage – final voltage)/peak voltage×100]. The PIAD waveform varies considerably from a standard monophasic capacitor discharging waveform. Critically, it has an almost negligible tilt (5%) compared with the standard (70%) or the rounded (60%) monophasic capacitor discharge waveforms (Figure 2).

Tilt reduction has been shown to result in lowering defibrillation threshold. Tilt was varied in these studies
by changing the pulse width of the waveform. Tilt can also be varied by changing the capacity of the energy-storing capacitor. An optimum pulse width is, however, required for successful cardioversion. The PIAD is capable of delivering a low tilt waveform regardless of the pulse width, enabling coexistence of an optimum pulse width and tilt.

**Waveform Rise Time**

The defibrillation threshold for capacitor discharge waveforms is suggested to follow an inverse strength–duration relationship. The average current at the shock amplitude providing 50% success gradually decreases with increasing pulse width, approaching rheobase. A mathematical model for defibrillation on a parallel resistor-capacitor network to mimic the heart was thus proposed, describing the response of the heart as a low-pass filter. Therefore, slow-rising waveforms should have an improved efficacy over acutely rising waveforms.

The slow-rising and -falling PIAD waveform may therefore have contributed to the efficacy observed in these studies.

**Waveform Rounding**

Waveform rounding has been shown to significantly reduce the delivered mean peak voltage and current compared with the standard while maintaining efficacy: the authors postulated that patients therefore would better tolerate an efficacious shock. This clearly is important for the functioning of an implantable atrial defibrillator.

Our findings using our rounded waveform demonstrated improved success rates for both the 50- and 100-V settings compared with the standard (Figure 6). The PIAD waveform, however, was more efficacious (phase 2) than the corresponding voltage setting for the standard or the rounded monophasic waveforms.

This is most likely helped by the waveform produced by the PIAD, which has a rounding of both the leading and trailing edges.

**Clinical Implications**

We envisage this novel device to be used in patients who have symptomatic paroxysmal AF that is not under pharmacologically or where pharmacotherapy is contraindicated and reversion to sinus rhythm is especially important. The PIAD does not have an implanted battery or a discharging capacitor and hence would not require replacement because of battery decay. This clearly will benefit patients, especially the elderly. The good success rates achieved at low voltage, current, and energy may be tolerated by patients with minimal or no sedation.

Patients with AF are not near syncope or syncopal during shock delivery. An automated device would cause chest discomfort especially if triggered during sleep and may be unsafe if triggered during driving. The PIAD can be activated only by the external transmitter. The lack of automaticity means that there is no risk of inadvertent shock delivery because of inappropriate AF detection. Furthermore, the absence of a complex algorithm analyzer, battery, and capac-

**Conclusions**

The PIAD, delivering a novel monophasic waveform, is an efficacious and safe method for cardioversion of AF. One hundred percent success was observed for the 10-ms 100-V setting (mean energy, 1.54±0.02 J) in a sheep model. The PIAD waveform was more efficacious than the standard or rounded monophasic waveforms. Furthermore, it is perhaps the waveform morphology (voltage and current profile) that predicts outcome of cardioversion rather than the mean energy delivered. This method provides another option in managing patients with symptomatic, drug-refractory paroxysmal AF, particularly the elderly.

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**References**


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