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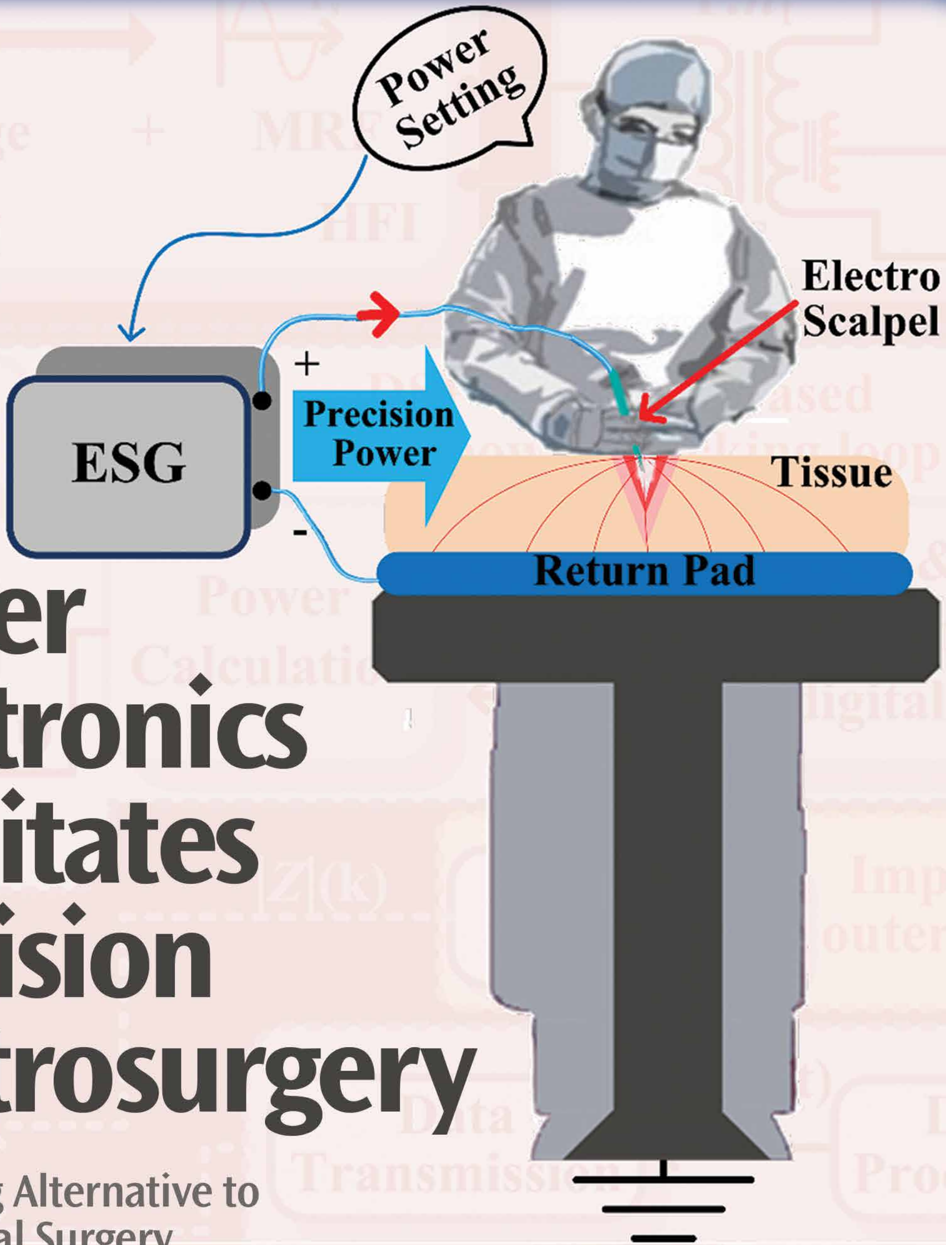

Power Electronics Facilitates Precision Electrosurgery

A Promising Alternative to Conventional Surgery

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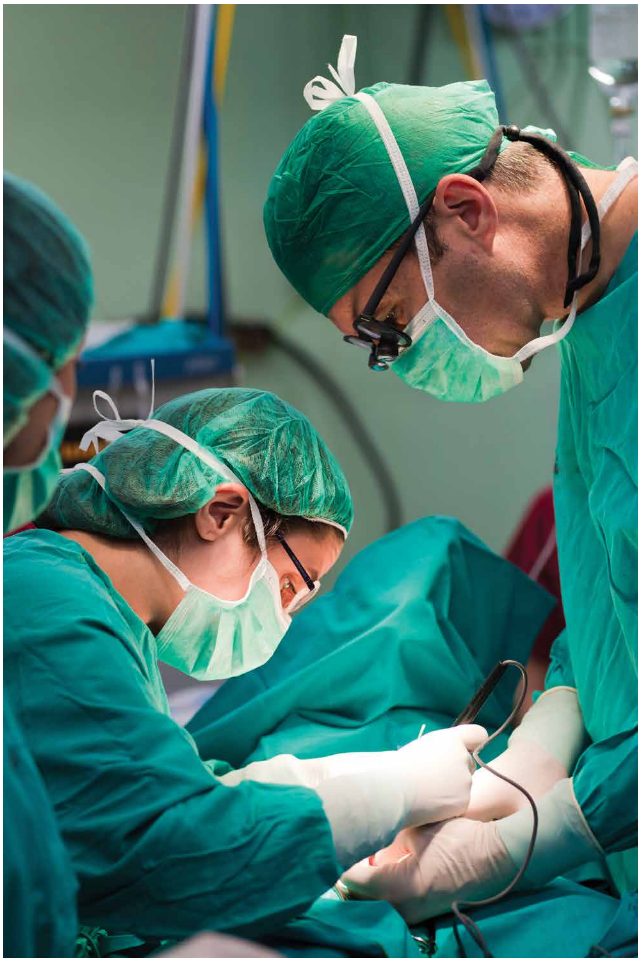

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Power-Electronics Enabled Precision-Power Electrosurgery

by Sudip K. Mazumder, Congbo Bao, and Ankit I. Mehta

Overview of Electrosurgery

In 1881, Morton found that a high-frequency (100-kHz) current could pass through the human body without inducing pain, spasm, or burn [1]. This work was followed in 1891 by d'Arsonval who noted similar observations at a frequency of 10 kHz and noted that the current directly influenced body temperature, oxygen absorption, and carbon dioxide elimination, increasing each as

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the current passed through the body [2], [3]. In 1897, Nagelschmidt identified [1] that patients with articular and circulatory ailments benefited from the application of electrical currents, which led to him proposing the term diathermy to describe the heating effect discovered in [2]. In 1900, Rivere, while treating an insomniac patient with electricity observed that a spark arcing from an electrode coagulated an area of his skin and subsequently, used this arcing current to treat a carcinomatous ulcer on the hand of a patient. This event has been cited as the first true use of electricity in surgery [1], [3], [4], [5]. Notwithstanding, many credit the invention of electrosurgery devices in 1926 to William Bovie [6].

Things started to change around one century ago when pioneers figured out a radically new way, namely, electrosurgery, to conduct surgery, which enriches the modality of biomedical surgery.

Modern electrosurgery passes alternating signals with a fundamental frequency above 0.2 MHz but lower than 5 MHz, through the human body to conduct clinical treatment, such as cutting, coagulation, and fulguration, etc., as illustrated in Figure 1. Such high-frequency (HF) current through the tissues of the body raises intracellular temperature to achieve vaporization or the combination of desiccation and protein coagulation thereby yielding a controllable surgical effect [7], presumably guided by exploding cell theory [8] or microscopic bubble theory [9].

The pure cutting mode, as shown in Figure 1, demands continuous electrical waveforms with a relatively lower amplitude [10], [11], [12], [13]. The tissues are vaporized with the utilization of a blade electrode during the cutting mode. In contrast, the coagulation mode requires a pulsed waveform with a higher amplitude [13]. The tissues, in the coagulation mode, are ionized by the applied high voltage and cool down during the blank period of the electrical pulse. Thus, the overall temperature of the tissue does not reach the vaporization point, leading to the coagulation surgical effect. More recently, some researchers have been working on continuous low-voltage-based soft-coagulation mode that dilutes the current density using a ball-shape electrode [14] and achieving coagulation by contacting the electrode to the biotissue. Finally, the blend cut lies in the middle with an adjustable period with output signals.

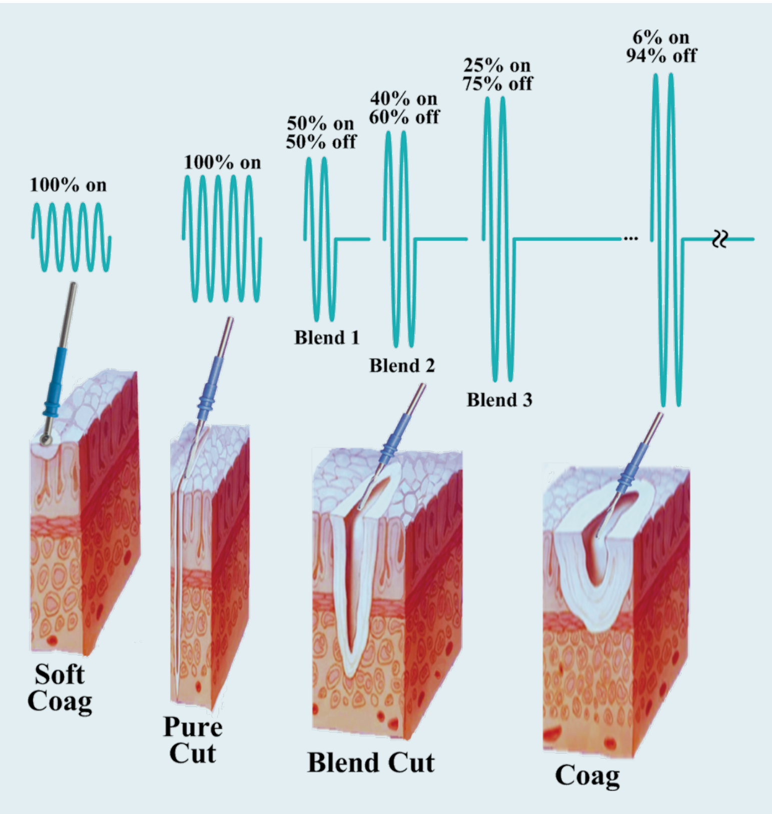

FIG 1 Illustration of some electrosurgery modes capturing electrical waveforms for cutting, blending, and coagulation and their physical manifestations.

Power-Electronics Enabled Electrosurgery

In conventional surgery, surgeons generally exert physical force on sharp stainless-steel scalpel for biotissue separation and utilize sutures for wound sealing or hemostasis. Things started to change around one century ago when pioneers figured out a radically new way, namely, electrosurgery, to conduct surgery, which enriches the modality of biomedical surgery [1], [15]. Unlike the conventional case, electrosurgery dissects the target biotissue by controlling the delivered energy to it through an "electro-scalpel" energized electrically by a high-frequency inverter (HFI), as depicted in Figure 2 [10], [11], [12], [13], [16].

In recent times, dedicated electrosurgery generators (ESGs) have evolved considerably from their older versions in terms of size, weight, functionality, galvanic isolation, protection for safety, attachment detection, and in multitude of other aspects [1], [17], [18], [19], [20], [21], [22], [23]. Furthermore, modern ESGs have evolved into two categories—monopolar and bipolar—depending on biotissue involved to complete the electrical circuit, power level, achievable bio-effects, etc. [15], [24]. However, the focus in this article is solely on monopolar ESG.

In order to transfer energy to a biomedical tissue for the expected surgical effect, an HFI is imperative.

Power electronics plays a significant role in the advancement of ESGs, which, in turn, is a key enabler for electrosurgery. With the development of power electronics, modern ESG has been equipped with various functions that support a wide range of surgical effects including but not limited to cutting, coagulation, desiccation, fulguration, and spray. Power electronics has also enabled the progressive miniaturization of ESGs in terms of size and weight [19], [20], [21]. The advent of wide-bandgap (WBG) semiconductor (e.g., GaN, SiC) based higher-switching-frequency and reduced-loss power electronics has further facilitated miniaturization of the ESG [16], [25], [26], [27].

In order to transfer energy to a biomedical tissue for the expected surgical effect, an HFI is imperative. Instead of HFI that generates a square-wave output voltage, one that generates a sinusoidal output voltage is preferred due to the reduced voltage slew rate, radiated electromagnetic interference (EMI), and magnetic losses when connected to a follow-up transformer for voltage step-up and galvanic isolation [10], [16], [28]. Furthermore, the output-voltage super harmonics that may inflict collateral tissue damage are diminished in a sinusoidal-output HFI [10], [23]. Consequently, a larger body of existing research or commercially-available ESG generate sinusoidal outputs at high frequency [10], [15], [16], [17], [20], [21], [22].

The frequency of an ESG output voltage much lower than 0.2 MHz may cause strong muscle or nerve stimulation, leading possibly to associated pain. On the other hand, for an ESG output frequency much higher than 0.2–5 MHz, cells may fail to polarize, and thus, the effects of the applied electric field on the cellular membranes may be considerably suppressed [19], [29]. As such, for a typical ESG, the output frequency ranges between 0.2 and 5 MHz.

Such a requirement essentially precludes the utility of typical pulse-wide-modulated (PWM) inverters, which have a wide temporal separation between the modulating and the carrier signals. This is because, to generate an ESG fundamental sinusoidal output in the 0.2–5 MHz range, the switching frequency of the ESG HFI may need to be in the tens of MHz range. From efficiency, thermal, and electromagnetic interference standpoints, operating at such high switching frequencies without complex and wide-range soft switching is potentially impractical.

Fortunately, many resonant HFI topologies exist, such as LCLC, MRF, class D, E, F, EF_n, Φ_2, and their variants [10], [13], [16], [30], [31], [32], [33]. These HFIs either generate a square waveform and reshape it to sinusoids by passing it to a resonant tank, or directly utilize or compensate device parasitics such that sinusoidal output is achieved while maintaining the device switching frequency the same as output frequency. The voltage profile across the semiconductor device in the former case is usually of square shape, however, it is a modified version in the latter one. It is possible to achieve soft switching in both cases for enhanced efficiency by reducing the device voltage or current to zero before switching transitions [34], [35]. However, attaining soft switching of most of those resonant HFIs suffer from certain limitations, such as high load dependence and device stress, complex printed-circuit-board (PCB) layout requirement, narrow load-supporting range, sophisticated tuning processes, and so on [10].

The full-bridge and multi-resonant frequency filter (MRF)-based HFI that enables electrosurgery with a fundamental (sinusoidal) output frequency of 390 kHz (with the inverter switching frequency being 390 kHz as well) is captured in Figure 3. The detailed working principle of this HFI is covered in [10], [16], and

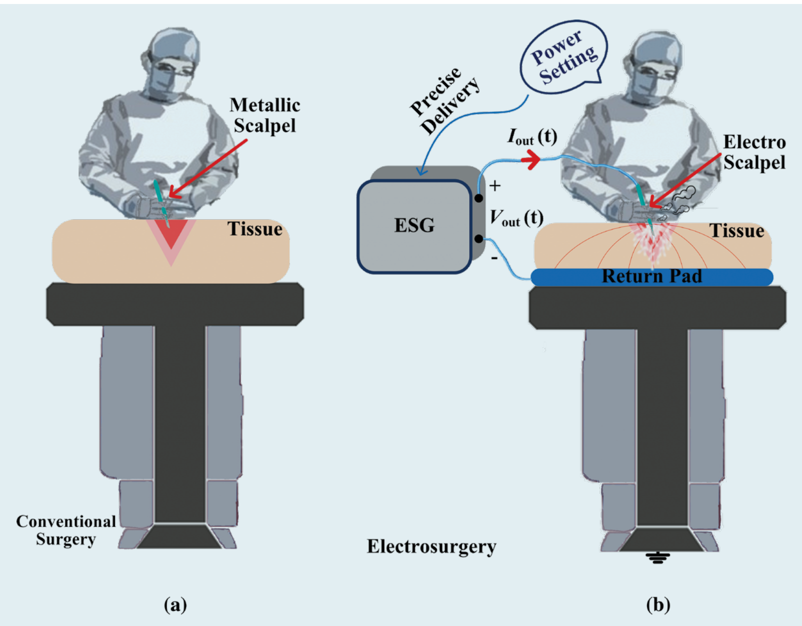

FIG 2 Illustration of (a) conventional surgery using metallic scalpel and (b) power-electronics-based electrosurgery generator using an electro scalpel.

[28], and its electrical characteristics are briefly described herein while its electrosurgical effects are narrated in the section below.

The device voltage stress in the MRF HFI is dictated by the magnitude of the input voltage while the output of the full-bridge HFI operating with phase shift control is a 390 kHz bipolar tri-state square waveform. The series resonant tank consists of C_1 and L_1, and it can be tuned to act as a short circuit at any targeted output frequency (390 kHz herein). The shunt resonant tank formed by C_2–C_4 and L_2–L_4 is in parallel with the transformer, and they are tuned at odd-order harmonics to be eliminated (third, fifth, and seventh harmonics here). The voltage division between two resonant tanks at different orders of harmonics is load-independent. Therefore, this MRF-HFI topology can support a larger range of loads with the added benefits of low output distortion, making it a good fit for electrosurgery applications.

Experimental results validate that both feedback mechanisms can automatically adjust the output power delivered to tissues experiencing electrosurgery.

Precision Power Electronics for Reduced Electrosurgery Tissue Damage

Beyond the power delivery for electrosurgery using a HFI, another goal is to achieve precise power delivery, manually set by experienced surgeons, to the target tissue per clinical needs. With the help of fast analog-to-digital converters (ADCs), properly designed output-sensing circuitry, advanced control design, and high-speed programmable processor (DSP or FPGA), it is feasible to obtain precision-power control via a power-tracking control loop.

But it is worth noting that the power precision is grounded on the high output-signal sampling rate allowed by fast ADC, especially, for the cases where nonlinear electrical arcing occurs in ESG outputs due to air breakdown owing to high electric field. The more samples that are captured per cycle, the more the digital-computation power is necessary to handle the data for HFI precision output-power generation. Consequently, a balance needs to be ensured between available digital throughput and desired power precision.

It is reported in the literature that power-control accuracy comes to a resolution of 1 W in some commercially-available ESGs. The variation in electrical impedance is dependent on the part of the body, or on individual patients requiring electrosurgery to a large extent. Therefore, the fixed power set by surgeons causes considerable collateral tissue damage in many cases.

One way of addressing this issue is to adapt the power delivered by the HFI to the electro-scalpel based on a real-time-feedback mechanism that can monitor the tissue status or reflect the quality of induced tissue effects, as depicted in Figure 4. By doing so, the modern ESG can achieve better clinical surgical effects with reduced or minimal collateral tissue damage.

One such feedback mechanism for precision power to reduce collateral damage, for instance, is to use thermal information where an infrared (IR) thermal sensor is mounted together with the electro-scalpel. The IR sensor continuously monitors the surgical-site temperature to generate real-time power adaptation as an outer feedback control loop such that the setpoint of the inner loop (power-tracking loop) is dynamically modified for an electrosurgery HFI [11], [13].

Another such mechanism, reported in [12], exploits measured information of tissue impedance as outer-loop feedback and performs power adaptation (or setpoint adjustment) for the inner-tracking loop in a real-time and cycle-by-cycle (390 kHz) manner. The load impedance-based approach has some superiority since it precludes the need for the thermal sensor and eliminates the corresponding drawbacks or issues related to it. Such drawbacks, to name a few, include slow temperature updates, the negative impact of smoke at the surgical site, added thermal sensor cost, and communication requirements.

Experimental results validate that both feedback mechanisms can automatically adjust the output power delivered to tissues experiencing electrosurgery. This also leads to reduced collateral tissue damage, as illustrated in Figure 5 [11]. All of these merits, together with versatile functions, make precision-power electrosurgery enabled by feedback-based power electronics potentially a radical departure from conventional surgery.

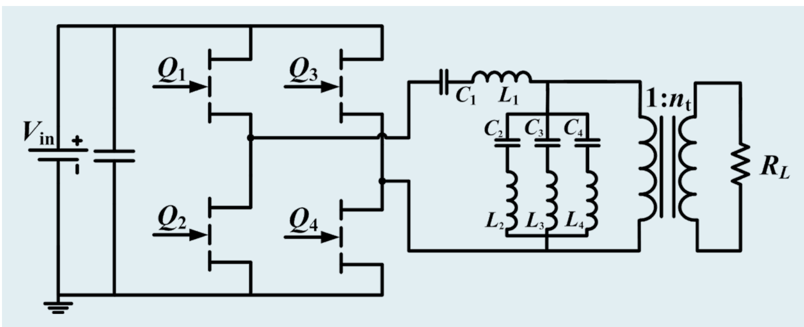

FIG 3 The topology of the full-bridge and MRF based ESG. The biomedical load is represented by R_L whereas V_{in}, Q_1–Q_4, C_1–C_4, and L_1–L_4 represent dc input, GaN devices, resonant capacitors, and inductors, respectively. The transformer for galvanic isolation and voltage boost-up with a turn ratio of n_t is articulated as well.

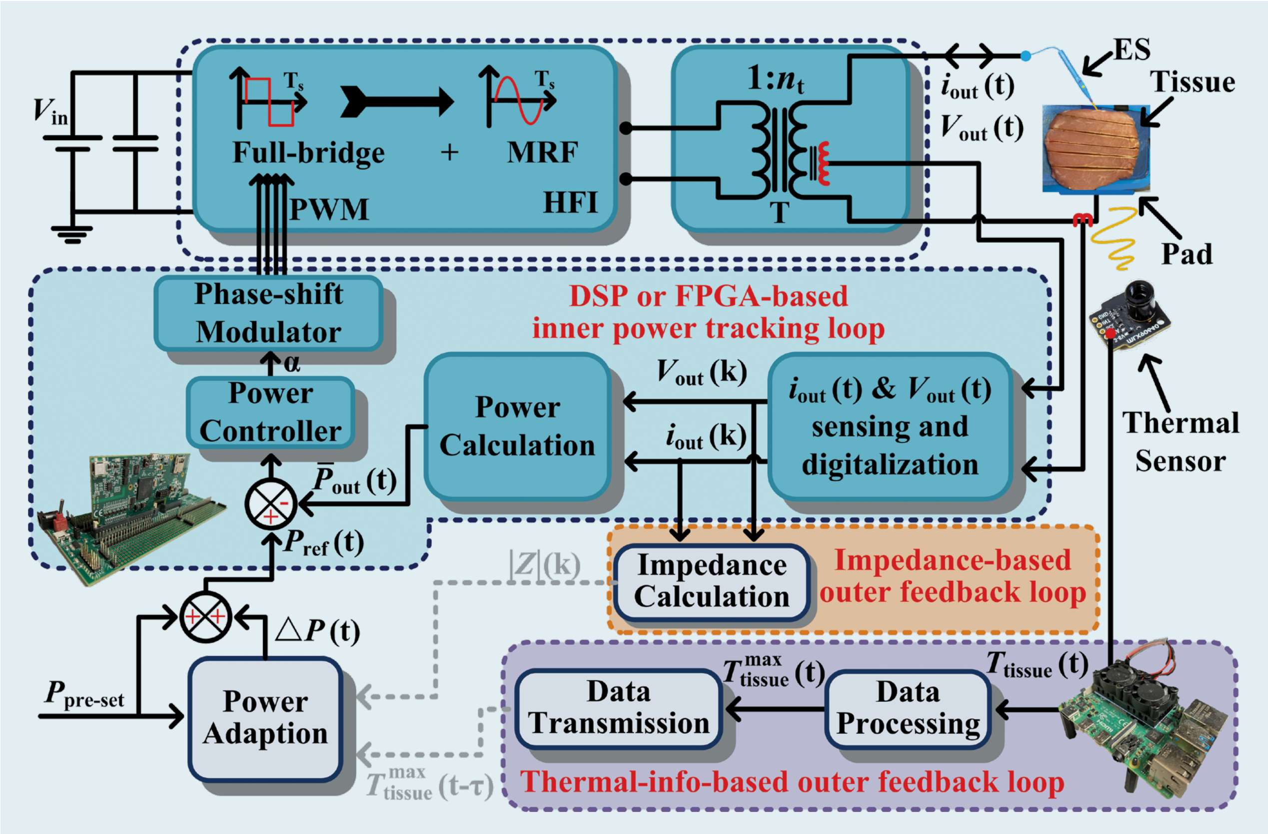

FIG 4 Illustration of temperature- and tissue impedance-based outer feedback loop that autonomously modifies the setpoint of the inner power tracking loop.

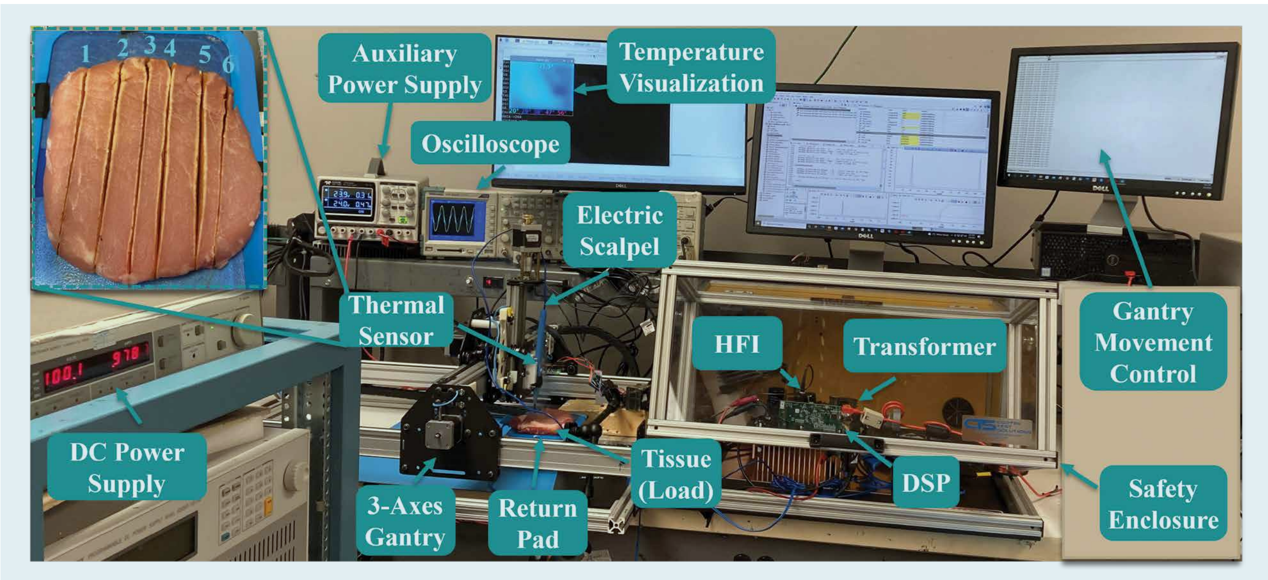

FIG 5 A HFI-based experimental test bed setup for electrosurgery. Trace 3 in the upper left picture illustrates how precision power yields reduced collateral damage.

Conclusions and Future Work

Power-electronics based electrosurgery is a promising alternative to conventional surgery for several surgical applications. A key technology in this regard is an HFI that needs to be carefully designed and operated. In that regard, given the high switching and output frequencies of the HFI, emerging wide-bandgap devices such as GaN FET is expected to play a key role. Yet another key aspect of electrosurgery is the synthesis of adaptive control (e.g., for power adaptation) that reduces collateral damage to the tissue. This article points to two recent approaches related to thermal sensing and impedance estimation that leads to adjustment of a preset fixed power (as typically set in conventional electrosurgery) thereby potentially reducing tissue collateral damage. With the advancements in computation, memory, sensing, and estimation one may expect further granularity in control and reduction in collateral damage. Finally, with regard to future work, the team is currently working on extending the advancements achieved so far to added functions of electrosurgery including coagulation, cutting and/or coagulation, soft coagulation, blending, etc. Yet another work is focused on extending the current work for detailed in-vivo experimental validations.

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