Radiofrequency (RF) energy, the dominant energy source for the treatment of various arrhythmias by catheter ablation, has replaced most other treatments for the effective suppression of supraventricular and ventricular arrhythmias.1 With expanding indications, it has become increasingly clear that the success and sustainable effect of RF ablation depends on a critical understanding of the biophysics of lesion creation and its control. RF current at the electrode tissue (ET) interface is the key to RF lesion creation by virtue of its tissue heating capability2 but cannot be measured directly, and therefore, delivered RF power and duration are typically manipulated to control lesion creation.3 Despite such manipulation, there is significant variability in lesion size, which is thought to be responsible for both inefficacy and complications. Contact between electrode and tissue may be the key parameter to control lesion size because it is essential for the passage of current into the target tissue, but has until recently remained intangible and unmeasurable.

Role of Contact in Lesion Formation

RF energy delivery creates tissue coagulation and necrosis by producing targeted tissue hyperthermia. This hyperthermia is the consequence of resistive heating of a thin rim of tissue in direct contact with the RF energy delivering electrode, with surrounding tissue being heated by conduction. Direct resistive heating is produced by a high density of alternating current, the magnitude of which declines with the inverse of the square of the distance; therefore, target tissue contact is essential for effective lesion formation.4 The key to effective and safe treatment with RF catheter ablation is controlling lesion size, that is, increasing or decreasing lesion size as desired. Precise control of lesion size is particularly important for treatment strategies requiring multiple coalescent ablations, for example, linear lesions making strategies, such as for atrial flutter, atrial fibrillation (AF), and reentrant ventricular tachycardia.4 The real-time measurement of ET contact may be an effective lesion size controlling parameter in the absence of direct measures of tissue interface current or tissue temperature distribution.5

What Is Contact?

ET contact can be considered to have 2 components: the magnitude of the surface area of the electrode in direct contact with tissue (contact footprint) as well as its stability.

Normal cardiac tissue like other soft tissue is both compliant and elastic: pushing a catheter in contact with tissue embeds the metal electrode tip progressively so that a greater percentage of the electrode surface is in direct contact with the tissue, that is, the contact footprint area increases. Spatial stability of contact refers to minimizing sliding of the tip electrode over the endocardium, whereas temporal stability of contact refers to maintaining stable levels of contact over time: avoiding both intermittent contact as well as excessive fluctuations in degrees of contact, typically because of respiratory and cardiac motion. Maximizing the contact area footprint reduces the electrode surface area exposed to the luminal circulating blood pool, that is, the low impedance shunt, thus favoring current delivery to target tissue5 (Table 1). Poor contact (temporal and spatial instability) can lead to current loss into the blood pool, coagulum generation, and failure to achieve appropriate myocardial temperatures, despite using adequate voltage and power.6

Pushing a catheter to achieve contact with tissue exerts a force on the tissue (and vice versa on the catheter). Force can be defined as any influence that can cause an object with mass to change its velocity (which includes to begin moving from a state of rest), that is, to accelerate, or a flexible object to deform, or both. The formal SI units for force are Newtons (N) or mN (milli-Newtons), but grams (g) are frequently used instead. Force is a vector quantity that has both magnitude and direction. Normal force (N) is a contact force that is perpendicular to the contact area, opposing separation of the surfaces (or pull of gravity across normal to the surface). Normal contact force is measured by the load cell mounted on the catheter and is the force per unit area applied in a direction perpendicular to the surface of an object. Thus, to derive the pressure exerted by the electrode tip on tissue (and vice versa), the normal contact force (CF) is used predominantly as a surrogate of the contact area or surface.
Real-time measurement of CF can allow evaluation of temporal stability, whereas spatial stability of contact may be better judged by ET imaging/localizing technologies.

Evaluating Contact

The earliest experiments described 2 different approaches to evaluating contact, one quantitatively measuring CF/pressure using a fulcrum and counterweights, whereas the other relied on qualitative visual estimation.

In the earlier study, RF lesions were created on canine right ventricular free-wall preparations with power adjusted to maintain a constant electrode temperature at 80°C. The electrode was positioned perpendicular to the myocardial surface, with contact being varied by appropriate counterweights (0–40.8 g) on the electrode support. Contact was measured as the force (in SI units, Newtons) exerted by the weights pushing the electrode on to the tissue. The authors observed that with increasing CF, the electrode deformed the endocardium to embed below the plane of the endocardial surface. The lesion width and depth increased significantly with higher CFs between ≈1 g and ≈10 g (and thereafter nonsignificantly up to 40 g), using power settings titrated to maintain a constant ET interface temperature. The absolute magnitude of increase in lesion size was small because of downregulation of delivered power to maintain ET interface temperature. This was probably the first use of the term CF in this context.

In another study, a catheter holder was used to position a 4-mm ablation electrode at 4 qualitatively different levels of contact (+3 contact or +1 contact—electrode pressed 3 mm or 1 mm into epicardium, respectively/0 contact or −5 contact—electrode lightly pressed or retracted 5 mm, respectively) on the epicardial surface of sternotomized anesthetized dogs. Lesser contact (with minimal catheter tip embedding or no tip contact) was associated with significant increases of power to maintain a constant ET interface temperature, and without contact 1 mm above the endocardium, lesion size was dramatically reduced because no direct resistive heating of the myocardium occurred. Lesser power was required to maintain the target temperature when the tip electrode was deeply embedded in the epicardium. The precise relationship between increasing contact and lesion size could not be directly evaluated because of the qualitative measurement of contact, but these observations clearly show that lesser contact results in smaller lesions with regards to volume and depth.

Contemporary Assessment of CF

Until recently, electrophysiologists have relied on tactile feedback, fluoroscopic monitoring of catheter shaft and tip movement, electrogram characteristics, and intracardiac ultrasound to estimate contact before delivering RF energy. Tactile feedback is subjective and difficult to titrate, whereas fluoroscopic monitoring is similarly nontitrable and requires X ray exposure. Electrograms only provide transient contact information and have been shown to be imprecise in judging contact. Ultrasound may hold out the promise of visualizing the catheter tip and tissue but microbubbles during irrigation and inherent resolution limitations are significant drawbacks.

Technological advances have now allowed the measurement or estimation of contact with otherwise conventional (irrigated) ablation catheters. Different technologies have been used in humans to provide continuous real-time estimation/measurement of tissue contact and CF (Table 2).

A robotic catheter contact sensor system (IntelliSense® Fine Force Technology) uses 2 force sensors that grip the shaft of any ablation catheter at the proximal end of the robotic system’s steerable sheath (Artisan). The ablation catheter is pulsed (<1.5 mm) in and out of the Artisan sheath (jittered), and CF is calculated based on mechanical resistance to movement with each pulse. An in vitro study examined the direct impact of catheter CF on lesion formation with this technology using intracardiac ultrasound and fluoroscopy during atrial ablation in canines. Echo-visualized qualitative catheter tip/tissue contact was correlated with CF values. The authors concluded that lesion size is optimized by the application of 10 to 20 g of CF and that mapping requires lower CF to avoid distortion and artificial increases in chamber volumes. A more recent in vitro analysis with this system also found a correlation between transmurality and CF, as well as an increased risk of steam pop and char formation with CF ≥40 g. The accuracy of CF measurement with this technology is thought, however, to be significantly angle-dependent and is temporally limited to 4 Hz. Furthermore, parasitic frictional forces (jitter) may be felt during catheter manipulation.

An impedance-based index (EnSite Contact VeriSense™) attempts to extract the local measure of impedance (unlike the standard global measurement between the catheter tip and a ground patch) specific to the catheter tip-to-tissue interface. The local coupling information is then provided as a real-time curve of electric coupling index (ECI) units. The index requires scaling to the individual patient, which needs to be repeated every 30 minutes to adapt to shifts in the baseline values during the procedure. Measurement and validation of the CF surrogate ECI were performed in the left atrium (LA) in patients undergoing AF ablation compared to unambiguous qualities of contact as determined by fluoroscopy, tactile feedback, and electrograms. A prospective randomized pilot study demonstrated the added value of ECI for lesion deployment as measured by higher rates of pulmonary vein isolation (PVI) after anatomic encircling. However, ECI and the nature of relative changes during tissue contact are not fully understood and remain more complicated than physical measures of CF. Individual scaling and the requirement for frequent calibration makes it difficult to compare different patients. Furthermore, ECI lacks precise distinction between different

<table>
<thead>
<tr>
<th>Table 1. Effects of Increased Contact Force</th>
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</thead>
<tbody>
<tr>
<td>1. Increased electrode–tissue interface surface area</td>
</tr>
<tr>
<td>2. Reduced electrode surface area exposed to (low impedance) blood</td>
</tr>
<tr>
<td>3. Reduced electrode tip sliding</td>
</tr>
<tr>
<td>4. Tissue compression and thinning</td>
</tr>
<tr>
<td>5. Tissue trauma</td>
</tr>
<tr>
<td>6. Higher tissue temperatures during RF delivery</td>
</tr>
<tr>
<td>a. Higher probability of pop</td>
</tr>
<tr>
<td>b. Higher probability of extracardiac heating</td>
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</table>

RF indicates radiofrequency.
levels of contact, as well as short instantaneous changes, for example, during cardiac cycle and respiratory movements.\textsuperscript{14}

The original catheter-based real-time CF sensor (TactiCath\textsuperscript{\textregistered} by Endosense–St Jude, Switzerland), incorporated within the distal segment of an open irrigated-tip (IT) ablation catheter, consists of a deformable body (elastic polymer) and 3 circumferentially aligned optical fibers. Any force on the catheter tip causes microdeformation of the elastic polymer, which is detected by a change in reflected wavelength off Fiber Bragg gratings within the optical fibres (and in the newer iteration by white light interferometry, also through 3 optical fibres). This signal is transformed to changes in CF (after suitable calibration in a free luminal position) with a sensitivity of 1 g and a sampling rate of 50 Hz (with the current white light interferometry technology).\textsuperscript{15,16} CF is calculated as a vector sum, thus providing the total force and its direction (ie, total force vector and its lateral and axial components).

A contemporary ex vivo model confirmed the importance of CF during RF ablation with an IT ablation catheter incorporating this CF sensor (see below).\textsuperscript{15} This study demonstrated that when controlled for power and duration of open-irrigated RF energy delivery, CF is the main determinant of tissue temperature and lesion size. The largest lesions were obtained using a high fixed power and CF (30 W/60 g). A higher CF was also associated with a higher tissue temperature, and a higher incidence of thrombus and steam pops. A later study confirmed these findings and also showed that the initial impedance at the start of the application and the impedance drop in the first 5 s correlated well with CF, suggesting the potential use of these parameters when direct CF measurements are not available.\textsuperscript{17,18}

We quantified the temporal evolution of CF (temporal stability) by calculating the force–time integral (FTI in gram-seconds, gs) or the area under the CF curve (Figure A), and correlated it in-vitro to lesion size, whereas RF power and peak CF were maintained constant.\textsuperscript{15} An open IT catheter incorporating a CF sensor was attached to a motorized movable mount programmed to simulate the movement of a beating heart. RF energy was delivered during intermittent contact (with loss of contact in diastole; Figure B), constant contact, and variable contact (without loss of diastolic contact; Figure C) for 60 seconds. The FTI was highest with constant contact, lowest with intermittent contact, and intermediate with variable contact. Furthermore, FTI correlated linearly with lesion volume between the different contact groups. This study in effect implied an in vitro estimation of lesion size by combining real-time CF and duration of (constant) RF delivery. The results also indicate that an intermittent contact pattern should be avoided to reduce ineffective RF delivery and, eventually, conduction recovery. Respiratory movements may also have similar effects on FTI (Figure D) and, thus, on lesion size, which is why FTI over 1 to 2 respiratory cycles may be useful to estimate lesion size and RF delivery during apnea may be preferred.\textsuperscript{20}

The force-sensing capability of the SmartTouch\textsuperscript{TM} catheter is based on the electromagnetic location technology used in the CARTO 3D mapping System (Biosense Webster, Inc, MA). The catheter tip electrode is mounted on a precision spring, allowing a small amount of electrode deflection. A transmitter coil coupled to the tip electrode, distal to the spring, emits a location reference signal. Three location sensor coils placed at the proximal end of the spring detect movements of the transmitter coil, representing deflection of the tip electrode on the spring. Based on known spring characteristics, the respective force and its orientation are calculated from the change in location of the tip electrode with 40 Hz temporal resolution. Calibration from a no-contact-free luminal position is necessary for an accurate CF reading sensitive to 1 g.\textsuperscript{21}

Because catheter manipulation results in variable and possibly excessive forces being exerted against the walls of cardiac

<table>
<thead>
<tr>
<th>Technology</th>
<th>Sensor Position</th>
<th>Temporal Resolution</th>
<th>Validation</th>
<th>Calibration</th>
<th>Units</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mechanical resistance to catheter pulsing within sheath (Hansen Medical, USA)</td>
<td>Within extravascular segment of proprietary sheath</td>
<td>4 Hz</td>
<td>Using ICE, fluoro in canines</td>
<td>Required</td>
<td>Grams</td>
<td>Accuracy angle–dependent; best results perpendicular to tissue</td>
</tr>
<tr>
<td>Measure of catheter-tip tissue electric coupling derived from 3 terminal circuit model of global impedance (St. Jude Medical, USA)</td>
<td>None; 2 indifferent patches required</td>
<td>Not specified; real-time curve generated</td>
<td>Evaluation during human PV isolation</td>
<td>Zerowing and calibration required</td>
<td>Electric coupling index (ECI); also ohms</td>
<td>Influenced by tissue characteristics, tissue heating, and overall impedance fluctuation during procedure; orientation independent, nonmechanical parameter</td>
</tr>
<tr>
<td>Originally microdeformation transduced by FBG optical fiber sensor; currently, transduced by white light interferometry through optical fibers (St. Jude Medical, USA)</td>
<td>Distal catheter tip: juxtaposed to tip electrode</td>
<td>50 Hz</td>
<td>Bench, in-vivo animal, clinical</td>
<td>Zerowing required; auto-zero effective thereafter</td>
<td>Grams</td>
<td>Tip orientation–independent CF measurement (see text), sheath constraint effect possible</td>
</tr>
<tr>
<td>Electromagnetic transduction of tip electrode microdeflection (Biosense-Webster, USA)</td>
<td>Distal catheter tip: transmitter coil attached to electrode with 3 location sensors proximal to spring</td>
<td>40 Hz</td>
<td>Bench, in-vivo animal, clinical</td>
<td>Zerowing required</td>
<td>Grams</td>
<td>Tip orientation–independent CF measurement (see text), sheath constraint effect warning, electromagnetic interference warning</td>
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</table>

CF indicates contact force; ICE, intracardiac ultrasound; and PV, pulmonary vein.
chambers, we evaluated the forces required to mechanically perforate the walls of swine cardiac chambers ex-vivo with a fiber-optic CF sensor. This study found that (a) the force necessary to perforate was significantly lower in the right atrium and right ventricle compared with the LA and left ventricle, (b) perforation forces were significantly lower through fresh transmural right atrial free wall RF lesions than through adjacent healthy, unablated right atrial tissue, and (c) cardiac perforation with a catheter within a sheath is quicker and easier than without the use of a sheath, which prevents the shaft from buckling and dissipating the force delivered by the operator. A later similar study with an electromagnetic CF sensor catheter in open-chest animals confirmed most of these findings. Recently, mechanical catheter perforation of swine and human (transplant heart) atrial tissue with a mechanical force tester confirmed lower right atrial versus LA perforation thresholds, as well as lower perforation thresholds for fresh RF ablation lesions.

Although no direct comparisons have been described, both the fiber-optic and electromagnetic CF sensors have been evaluated (separately) on the bench against certified balances, and the measurements were found to correlate highly at 3 different angles. Both technologies provide comparable temporal resolution and indicate force vectors. However, there is little information about the dynamic resolution of their force measurements or the accuracy of high amplitude forces with the 2 technologies. Whether tip deflection versus microdeformation has differing effects on the relationship between force and the contact footprint also remains unknown.

Figure. A, Real-time contact force (CF) curve showing calculation of force–time integral (FTI; 183 gs) as area under curve (red hatched zone) during short duration of radiofrequency (RF) energy delivery. B, Real-time CF curve showing intermittent contact. C, Real-time CF curve showing variable near-constant contact. D, Real-time CF curve showing respiratory variation. Numbers on the left indicate mean CF (grams) and on the right indicates FTI (gs).
The study concluded that the safety profile was comparable to conventional IT RF catheters, but also highlighted a marked inter- and intraoperator variability of CF during subjective CF-blinded assessment of optimal contact (Table 3). High CF values (>100 g) were observed even during catheter manipulation, and the single perforation event in the study was preceded by significantly high transient force (137 g) during catheter manipulation, suggesting that avoiding CF values >100 g throughout the procedure is important for patient safety. Low CF values are also advisable in the vicinity of already ablated sites, where the tissue is structurally weakened. Although not a primary end point, all AF patients treated with a mean CF <10 g experienced recurrences, whereas 80% of patients treated with a CF of >20 g had none during follow-up. The FTI (gs) also correlated with freedom from recurrence.

The relationship between local CF during RF application and gaps at the LA–pulmonary vein junction was prospectively evaluated in the Efficacy Study on Atrial Fibrillation Percutaneous Catheter Ablation With Contact Force Support (EFFICAS I) study, with an invasive electrophysiological reassessment 3 months after index ablation. Because the study found a strong correlation between minimum CF and FTI values and subsequent gaps, the authors suggested that optimum CF is required before ablation to minimize ineffective lesion formation, targeting an average CF of 20 g, a minimum CF of 10 g, and a minimum FTI of 400 g.

The preliminary results from the TactiCath Contact Force Ablation Catheter Study for Atrial Fibrillation (TOCCASTAR) trial, a prospective, multicenter, randomized study comparing non-CF–based IT ablation with fiber-optic CF sensing IT ablation of paroxysmal AF showed noninferiority with respect to non-CF ablation. A subgroup with >90% of lesions delivered with CF >10 g showed superior rhythm outcomes. The SMART-AF study, a prospective, multicenter, nonrandomized study of paroxysmal AF ablation, found that an increased percent of time within physician-chosen target CF ranges correlated with increased freedom from arrhythmia recurrence. However, the chosen target ranges were variable, and possibly as a consequence, a higher CF did not correlate with effectiveness. A slightly higher than expected incidence of tamponade was observed, which could be attributed to efforts to increase CF and should perhaps signal a cautionary note.

Several single center acute or short-term studies have achieved PVI with less ablation time/lesions and lower rates of adenosine-provoked conduction recovery using CF catheters. One prospective study found more efficient procedural outcomes and better 1 year arrhythmia-free outcomes with CF sensing. Another acute study found that local FTI provided the highest sensitivity and specificity for effective PVI lesion creation. A single center study also showed that an FTI >1200 g during PVI correlated with larger late gadolinium enhancement areas on 3 month post PVI magnetic resonance imaging scans. In our laboratory, a minimum CF of 10 g is required for ablation targeting PVI or for linear atrial lesions.

**Catheter Stability and Other Factors**

Spatial catheter stability and CF during RF ablation can be improved with various adjunctive measures, such as general anesthesia or high frequency jet ventilation (to control
respiratory motion), or adjunctive tools, such as steerable sheaths. In addition to increased spatial stability, steerable sheaths allow the achievement of higher CFs, at least partly by increasing the column rigidity of the catheter-sheath combination. Higher clinical success for RF ablation of AF has been achieved without increased complications with a manually controlled steerable sheath for catheter navigation.36,37 Catheter tip tracking software may also facilitate greater spatial stability of contact; however, currently it is not possible to track the catheter tip position with respect to the heart (which itself exhibits variable respiratory and cardiac movements).38

The effect of respiration on CF during ablation in the cavo-tricuspid isthmus has been studied with apnea compared with ventilation, allowing the achievement of higher FTIs and more efficient acute achievement of complete conduction block.19 An innovative approach has tried to compensate for cardiac motion by gating pulsed RF delivery to electrograms with experimental evaluation showing deeper lesions with electrogram-gated RF delivery.39

Regional variations in CF are frequently encountered during mapping and ablation procedures21 and may be accounted for by differences in myocardial tissue composition, underlying organs or tissues (eg, compliant lung tissue versus the blood filled aorta) or variations in transmission of operator-delivered force (because of differences in column rigidity—eg, sheaths, or significant catheter angulations or other catheter delivery variables). However, it is unclear whether differing myocardial tissue composition or underlying structures would modify the CF-lesion size dose–response curve.

Lesion Modeling, Size Estimation, and Prediction

During RF delivery, local electrogram attenuation, electrode temperature, and impedance drop have been used as approximate indicators of effective lesion creation.9,32 Electrode temperature feedback from tissue heating is, however, largely neutralized or eliminated during open IT RF ablation, and the impedance measurements are sensitive to the many parameters that could affect any component of the typically large unipolar RF energy delivery circuit.

The ability to measure CF in real-time may significantly improve the accuracy of lesion modeling and prediction. The FTI can be considered an early iteration of clinically useful approaches to RF lesion estimation, assuming constant RF power and a limited RF delivery duration.19 Simplified approaches combining CF, RF power, and delivery time are being developed and have shown promising correlations with in-vivo lesions in animal models.40,41 A recent modeling study incorporated all the different variables relevant to lesion generation (including CF amplitude and vector orientation, which determine contact footprint) in mathematical terms, deriving an aggregate equation to describe RF lesion creation.42 When tested against in vitro bench RF ablation, the model predicted tissue temperatures accurately in near real time at multiple specific distances and depths over the course of RF delivery. In conjunction with sophisticated imaging and tissue interrogation (eg, high resolution computed tomography or magnetic resonance imaging), an effective lesion prediction algorithm could improve efficacy and safety by tailoring lesion size to the target tissue, its thickness, and surrounding vulnerable structures. We target an FTI >500 g to try to achieve transmural adequate-sized individual atrial lesions, typically with a steerable sheath and 30 to 35 W power. We think it is important to avoid both high powers and high CFs, particularly during the initial steeply rising portion of the curve-relating lesion size/depth to duration and therefore deliver RF for 25 to 60 s with moderate power and CF. We downregulate RF power when CF cannot be reduced to moderate ranges without catheter instability or in the presence of adjacent vulnerable structures.

Catheter CF for Mapping and Ablation of Ventricular Arrhythmias

CF parameters used predominantly for the ablation of atrial tissue have also recently been evaluated for ventricular mapping and ablation.

Mapping and ablation of ventricular tachycardia can be challenging. Apart from clinical parameters, such as multiple and nontolerable ventricular arrhythmias, recognition and optimization of electrode tip–tissue contact is just as crucial as in atrial tissue. The greater thickness of ventricular myocardium compared with atrial myocardium makes it more important to appropriately titrate lesion depth. The greater ventricular systolic-diastolic excursion (compared with atrial contraction) is likely to predispose to intermittent contact with loss of contact in diastole. The distinction of normal versus abnormal ventricular myocardial substrate (dense scar, border-zone, inhomogeneous scar, midmyocardial scar) is currently frequently made based on electrogram characteristics, such as voltage and the presence of late potentials. There is increasing evidence that CF parameters may significantly affect these parameters.

To date, 2 studies have addressed these issues with largely congruent results.43,44 Mizuno et al were the first to assess the use of CF-sensing catheters in ventricular tachycardia mapping in humans.43 They compared a combined antegrade, transeptal (with a steerable sheath), and retrograde, transaortic mapping strategy with a retrograde-only approach in 17 different patients, divided all acquired mapping points into 2 groups according to the presence of positive CF throughout a complete cardiac cycle (ie, also during diastole), and found that surrogate parameters, such as floroscopy, electrogram amplitude, and local impedance, were unsatisfactory for predicting/monitoring tissue contact. The authors advised a CF of 8 g for left ventricular and 9 g for right ventricular mapping. Low CF (<3 g) resulted in low amplitude electrograms, whereas high CF (>20 g) was not associated with any further increase in voltage when compared with moderate values (10 ≤CF ≤20 g). Falsely, low voltages because of insufficient CF may have significant implications for substrate mapping and voltage-derived definitions of scar tissue. Intriguingly, they also showed that abnormal late potentials could be frequently missed because of low CFs—in as many as 30% of sites. Tiltz et al recently added more evidence in line with these observations, during assessment of the impact of antegrade–transseptal

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with retrograde–transaortic left ventricular mapping on catheter stability and CF and surrogate parameters for tissue contact. Ventricular lesion formation after RF ablation on the right and left ventricular endo- and epicardium has been investigated using a standard IT catheter versus a CF sensing IT catheter. Conventional surrogates of good ET contact, such as fluoroscopy, tactile feedback, and electrogram amplitude, did not result in endocardial lesion formation in 22% of RF applications, whereas lesions were only absent in the CF-sensing group when RF was applied with a CF <10 g or an FTI of <500 g—in line with the results from the TOCCATA study. In our laboratory, we try to avoid delivering RF in the ventricles when the mean CF is <10 g and the diastolic CF is <5 g, despite the absence of corroborative data.

The preceding absolute CF parameters do not seem to be directly applicable to predict lesion size in epicardial RF delivery, where despite lower FTIs compared with the endocardium, lesion volume was significantly larger: most probably because of absence of circulating blood and lack of convective cooling. Although the presence of epicardial fat has a considerable modifying effect on lesion formation, the catheter also has a more parallel orientation on the epicardial surface, so that the applied CF has a greater lateral than axial component, resulting in altered lesion geometry. Such force vector information could be useful to prevent pulmonary lesions during epicardial RF ablation. These insights have recently been reported to be helpful for successful CF-guided endocardial and epicardial ablation of ventricular tachycardia.\

**Future Directions**

It is likely that multielectrode RF ablation (eg, for linear or circular lesions) would benefit from individual electrode-based real-time CF assessment of contact. Precise and accurate lesion prediction and estimation algorithms incorporating CF are being developed and will be validated. Electrogram voltage criteria for myocardial scars will likely be shown to have greater sensitivity and specificity when validated by quantitative CF criteria. The availability of real-time CF sensing may also provide a new impetus to the development of fully automated robotic catheter manipulation and ablation. Despite increasing investigation into real-time lesion visualization or evaluation, perhaps with ultrasound, magnetic resonance imaging, or other technologies, CF sensing is likely to retain a key role because it allows pre-RF delivery titration, provides useful information during manipulation, and remains an important parameter controlling lesion size (Table 2).

**Conclusions**

Catheter-based real-time CF measuring technology has provided important insights into the biophysics of lesion creation using RF energy. Adequate CF is fundamental for the acquisition of reliable mapping information and achievement of stable ablation lesions. The real-time measurement of CF is a clinically relevant technology that can help achieve accurate control of lesion size.

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