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Virtual 3D heart models to aid pacemaker implantation in children

Natalia A Trayanova[*]

[*]Department of Biomedical Engineering & Institute for Computational Medicine, Johns Hopkins University, Baltimore, MD, USA; ntrayanova@jhu.edu

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congenital heart disease; defibrillation threshold; device implantation; electric shock; finite element modeling; image-processing pipeline; implantable cardioverter–defibrillator; pediatric patient; virtual heart model

Defibrillation by a strong electric shock remains the only known effective way of terminating ventricular fibrillation and thus preventing sudden cardiac death. External defibrillators have long been used as standard therapy, and implantable cardioverter–defibrillators (ICDs), based on mature foundational technology that evolved from antibradycardia pacing, have also been demonstrated to be an effective, life-saving technology. Large, well-controlled prospective ICD therapy trials such as AVID, MADIT-I and -II, and MUST have revolutionized the concept of sudden cardiac death prophylaxis and have proven the efficacy of ICDs for both secondary and primary prevention of sudden cardiac death [1]. These studies have led to the rapid growth of patient populations for whom ICDs are indicated, with over 110,000 devices implanted annually in the USA alone.

As a result, increasingly diverse populations of patients have been undergoing ICD implantation. In 1989, the first use of ICDs in young patients was reported [2]. Since then, ICD therapy has become increasingly important as a treatment approach in the pediatric population [3], with the mean age at implant decreasing significantly (from 13.6 to 12.2 years), and the percentage of patients younger than 5 years of age receiving an ICD increasing to above 10% [4]. The clinical outcome of ICD therapy in pediatric patients and adults with congenital heart disease has been evaluated by large multi-center studies, which included more than 200 children [5,6].

Despite the increased ICD implantation rate and the successes in ICD therapy, children and patients of small body size, as well as patients with congenital heart disease, are poorly served by current ICD technology. In these populations, the procedural approach and site of implantation, therapeutic algorithms, and early and long-term complications are different from those in adults, and the optimal ICD implantation technique has not yet been established. Furthermore, transvenous approaches to lead implantation presents a problem because of the small venous system [7] – there are no specific electrodes for small vessel

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diameters – and because of the frequent congenitally altered heart anatomy. As a result, transvenous approaches are associated with an increased risk of venous thrombosis. Nontransvenous ICD systems used currently in clinical practice in younger patients and in those with smaller body surface area, intracardiac shunts, and concurrent thoracotomy surgery or affection of tricuspid valve include pericardial and subcutaneous coils. In addition, existing ICD devices are not adapted to the pediatric patient's small body surface and weight, frequently requiring abdominal implantation of the power generator. Early physical activity and impaired sterile conditions lead to more frequent procedural complications in these patient populations. Inappropriate discharges, lead-related complications and generator anomalies are the most common adverse events occurring during follow-up; Lewandowski *et al.* reported a 21% rate of complications requiring surgical intervention [

The variability and structural complexity of pediatric hearts, particularly those with congenital heart defects, make ICD device implantation and management a highly individualized art. There is currently no reliable, personalized way of predicting which ICD configuration would have the lowest defibrillation and/or cardioversion thresholds in such a patient. A wide variety of innovative approaches to ICD implantation have been demonstrated to be clinically feasible in children and congenital heart patients. These approaches have generally utilized 'off-the-shelf' technology in untested ways. Little data are available to guide the specific application of defibrillation strategies in these patient groups. Clinical studies in these patient groups to determine appropriate defibrillation parameters are difficult to design and complete due to ethical and practical considerations. The clinical data indicate that because ICD technology is deployed into clinical situations for which it has been neither designed nor extensively validated by clinical experience, there is now a pressing need for alternative approaches to technology development and therapy planning in the field of cardiac defibrillation, approaches that are both flexible and incorporate our increased knowledge of the mechanisms by which an electric shock defibrillates the heart.

By marrying clinical MRI with sophisticated computer analysis, the paper by Rantner *et al.* has now provided proof of concept that it is possible to take the guesswork out of the ICD implantation process in pediatric and congenital heart disease patients [9]. The researchers used an MRI-based patient-specific 3D heart–torso model that takes into account the child's unique heart anatomy, and determined the optimal locations (in terms of minimum shock energy) for both leads and the power generator before the device is implanted.

In the past, finite element modeling has been used by numerous investigators to model human defibrillation. Specifically, models of the human thorax have been employed to predict the intensity of an electrical field delivered by a defibrillator device, as well as the amount of current that reaches the heart during this process. Attempts to construct heart–torso models from MRI data sets have also been made previously, but only for patients with structurally normal hearts. Such models have been employed in the study of defibrillation [10–13], as well as for other uses [14]. However, all of these prior models were limited to predicting only the static thoracic electrical field induced by the electric shock, and relied on inference to predict the actual electrophysiological effect of this field on the fibrillating heart. To adequately describe this response, it is necessary to include in these models a biophysically accurate, mechanistic representation of the interaction of the defibrillation electric field with the myocardium.

Studies have demonstrated that the cells in the heart respond to the electric current delivered by the defibrillator device in a strongly nonlinear fashion, by generating concurrent regions of positive and negative membrane polarization throughout cardiac tissue [15,16]. This effect is termed ‘virtual electrode polarization’ (VEP) and represents the essence of the interaction between cardiac tissue and the applied electric field. VEP has been documented in numerous experimental studies and is supported by computer simulations [17–19]. Furthermore, research has shown that in addition to the nonlinear membrane response, cardiac tissue structure, particularly fiber architecture, plays an important role in the generation of VEP and its shape, location, polarity and intensity [20]. The distribution of VEP throughout the heart in response to the shock determines whether existing wave fronts in the heart are terminated by the shock, and whether new wave fronts are generated that could reinitiate arrhythmia. Clearly, accurate prediction of the generation of VEP and subsequent propagation of postshock activation is necessary for the accurate prediction of shock outcome, and requires explicit representation of both the electrophysio logical properties of the myocardium, as well as the myocardial geometry and fiber architecture.

The study by Rantner *et al.* presented the first electrophysiological, active multiscale heart–torso model that accounted for the aforementioned interactions between an applied electric field and the cells in the heart [9]. Importantly, the study developed a new image-processing pipeline for building patient-specific heart–torso models from low-resolution clinical MRIs, which was applied, in a proof of principle, to a pediatric patient with congenital heart disease. Using the pediatric patient heart–torso model constructed with this image-processing pipeline, the study used simulations of the defibrillation process to determine the shock outcome for different ICD configurations. Ventricular fibrillation was induced in the model heart, and defibrillation shocks were applied from 11 ICD configurations to determine the outcomes of shocks and the defibrillation thresholds, the minimum amount of energy that successfully defibrillated the heart with shocks from a given ICD configuration. Two configurations with epicardial leads resulted in the lowest defibrillation thresholds overall and were, thus, considered optimal. The study also demonstrated that in order to reliably predict defibrillation outcome, the patient-specific heart model needs to accurately incorporate the biophysically detailed interaction between the applied shock and the cells of the heart. The pipeline and methodology developed in this study presented a novel approach to predicting the optimal ICD configurations before device implantation, a tool of particular importance, as reviewed above, for pediatric and congenital heart disease patients who have contraindications for transvenous lead ICD implantation. The clinical translation of this approach could provide a reliable, personalized way of predicting the location of ICD device placement in such patient populations.

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