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Computerized Optimization of Biventricular Pacing Using Body Surface Potential Map

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Abstract— **An improvement of biventricular pacing (BVP) could be possible by detecting the patient specific optimal pacemaker parameters. Body surface potential map (BSPM) is used to obtain the electrophysiology and pathology of an individual patient non-invasively. The clinical measurements of BSPM are used to parameterize the computer model of the heart to represent the individual pathology. The computer model of the heart is used to simulate the dyssynchrony of the ventricles and myocardial infarction (MI). Cardiac electrophysiology is simulated with ten Tusscher cell model, while excitation propagation is intended with adaptive cellular automaton at physiological and pathological conduction stages. The optimal electrode configurations are identified by minimizing the QRS duration error of healthy and pathology case with/without pacing between pre and post-implantation. Afterwards, the simulated ECGs for optimal pacing are compared to the post implantation clinically measured ECGs. The optimal electrode positions found by simulation are comparable to the ones meausured in hospital. The QRS duration reduction error between measured and simulated 12 ECG signals are similar with a constant offset of 15 ms. The personalized model present in this research is an effective tool for therapy planning of BVP in patients with congestive heart failure.**

I. INTRODUCTION

Clinical trials of CRT have demonstrated considerable improvements in quality of life and exercise capacity, but a significant number of non-responders have decreased the overall benefits [1-3]. The use of ECG criteria alone might result in selection of some patients who are unlikely to benefit and also exclusion of potential responders. Assessment of the regional left ventricular mechanical activation and viability with the echocardiography should be considered before implantation of the BVP device and further studies are required to refine the selection process. Once a decision has been made to proceed with the CRT then appropriate placement of the leads and optimal programming of the device will maximize the therapeutic effect. A recent study by Mohindra et al. [3] investigated the use of body surface potential mapping and computer model simulations to optimize the CRT device. The outcome was the influence of pacing locations on reducing dyssynchrony. Placing the left ventricular (LV) pacing tip on the posterobasal section of LV and pacing both leads simultaneously helps to greatly reduce dyssynchrony without the need for optimal programming of the delay. This suggests that in patients with optimal lead placement, programming of the inter-ventricular (V-V) timing may not be necessary [3-5]. Ideally, a computerized optimization of BVP would use a model of the cardiac contraction in order to calculate the cardiac output. Since these models required an extensive calculation time and were not available at the time of this analysis, an alternative approach was used. If the temporal and spatial excitation propagation of the pathological heart with pacing is as close as possible to the physiological excitation propagation, it is assumed that the cardiac output will be optimal. Thus, the activation times for each cardiac cell (isochrones) are computed for physiological, pathological and therapeutic case (pathology with pacing). Therefore, the presented work proposes a non-invasive optimization algorithm to find the best electrode positioning sites and timing delays for BVP in patients with left bundle branch block (LBBB) and MI. This algorithm can be used to plan an optimal therapy for an individual patient. The optimization algorithm is applied once on the activation time and other time on ECG intervals to investigate the BVP parameter optimization.

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II. METHODS

The representation of the model generation including the relations between different parts is illustrated in Fig. 1.

The purpose of the work is classified in two strategies. The structure of the first strategy is:

- Construction of an individual patient heart model considering the LBBB and MI extracted from the segmented MR data sets according to the description of the excitation conduction system and pathologies.
- Assessment of the optimal pacing configurations as well as the best electrode set-up and timing delays in the course of two electrophysiologically based optimization algorithms.
- Comparison of the results obtained from the two optimization methods and reconcilement of the derived results.

The results of the two optimization algorithms may differ. In this case, a trade off between optimal results could be

performed, ensuring a proper selection of pacing parameters. The structure of the second strategy is:

- Creation of an individual patient torso model considering the inner organs such as heart, lungs, liver.
- Solution of the inverse problem with patient model parameters set to achieve a personalized model of the pathology by measuring the multichannel ECG from a patient prior implantation.
- Calculation of the BSPM for the patient under therapy based on the optimal BVP set-up.
- Extraction of the 12 standard ECG channels.
- Verification of the optimal pacing results with respect to comparison of measured ECG of the patient after implantation with the simulated ECG for optimal pacing parameters.

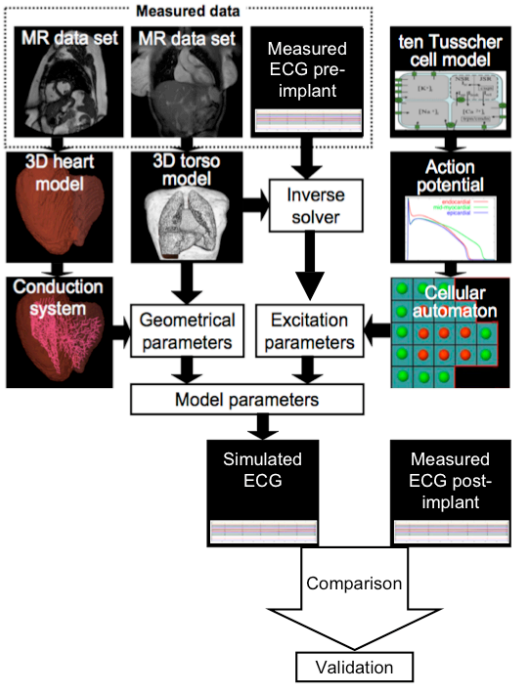

Fig. 1. Complete overview of model generation, from in-vivo measured data to the model parameters. The arrows are indicating the link between the segments and demonstrate the order of the computations. The pathology is implemented by altering the parameters in the conduction system and cellular automaton. The torso model jointed to the cellular automaton can be used to simulate the ECG, making possible direct comparison of measured ECG, enabling clinical validation.

The 3D volume conductor model of the thorax was extracted from a series of frontal axis 2D MR data sets for the patient. The tetrahedral mesh of the thorax was created including the heart and the other organs. The action potential (AP) of human myocytes was simulated by the ten Tusscher ionic cell model [6]. An adaptive cellular automaton (ACA), belonging to the rule-based heart models, was used in the present work. The ACA model does not consider explicitly the ionic flow interaction between the intra- and extracellular spaces in order to simulate the excitation propagation. Instead, the pre-calculated APs derived from the ionic current equations based on the ten Tusscher model are applied as a set of rules stored in a predefined library for the fast computation of ventricular excitation. To extract the optimal pacemaker set-ups including the electrode positions, atrio-ventricular (A-V) and inter-ventricular (V-V) delay, the optimization algorithms are applied on isochrones. The activation times of each cardiac cell were calculated for the physiological, pathological and therapeutic case. The divergence of the isochrones was calculated as a root mean square error (E_{RMS}). Where, E_{RMS} was taken as a measure for the difference between physiological and pathological excitation propagation. E_{RMS} was minimized in order to find a pacing set-up delivering a temporal and spatial excitation propagation as close as possible to the natural physiological state, assumed to produce an optimal cardiac output [7, 8].

Since the ECG of a patient with a LBBB was characterized by a QRS complex with the duration of more than 120 ms [9], the QRS duration was computed for all cases (1). The QRS duration was calculated as the difference of the activation time of the first and last activated cardiac cell. According to the selected optimization method, the optimal BVP parameters are those leading to minimal QRS duration difference between case 1 and 3 (2).

$$t_{QRS} = t_{act,last} - t_{act,first} \quad (1)$$

t_{act}: activation time of the corresponding voxel,

t_{QRS}: QRS duration time.

$$t_{error} = t_{QRS.path(pace)} - t_{QRS.phys} \quad (2)$$

t_{error}: time difference between physiological and pathological/pacing case.

The steps of the optimization algorithm are the following:

- Calculation of the QRS durations for the physiological and pathological case.
- Calculation of the QRS duration for pacing case.
- Finding the minimal difference of QRS duration between physiological and pacing case.

In total, 36 electrode set-ups were investigated per patient. Eight different electrode positions were chosen in the anterior (A, B, C, D) and posterior (I, J, K, L) branches of coronary sinus. Four positions were chosen in the left ventricular free wall (E, F, G, H) based on several studies [10]. The right ventricular electrodes were placed in the right ventricular apex (X), upper (U) and middle (M) septal wall (Fig. 2). In order to speed up the calculation process for timing delay optimization, a Downhill simplex algorithm (DSA) was utilized [8]. The measured ECG signals were processed with the following algorithms: denoising, baseline wander correction, averaging in time, R-peak and QRS detection. For post-implantation data, the stimulus removal algorithm was added to the signal processing. The removal of the baseline wander was performed with linear filtering, polynomial fitting and wavelet transforms. The wavelet transform was applied to the ECG signals in order to

decompose them to the low and high scales coefficients. The wavelet transform of the ECG signal was computed according to the method proposed by Tinati et al. [11].

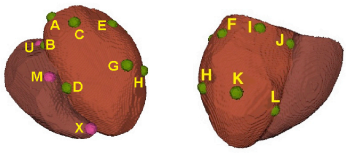

Fig. 2. The electrode positions chosen for the heart model.

In each scale, the wavelet coefficients energy for both coarse and detail levels were calculated (branches of the tree). Once comparing the energies of coarse and detailed levels, the branch of the tree with the higher energy was chosen. The higher energy branches were consecutively followed until was reached a point where the energy difference exceeded a preset threshold level. At this point the tree was completed, and the baseline wander signal was identified. Using the obtained wavelet coefficients, the inverse wavelet transform was calculated after subtracting the baseline wander from the wavelet coefficients. Thus, a baseline wander free ECG signal was obtained. In order to remove the noise from the ECG signal, discrete wavelet transform (Daubechies) was applied to the signal. A threshold was selected according to Donoho's thresholding method and then applied to the detailed wavelet coefficients [12]. Finally, the denoised ECG signal was reconstructed with the inverse discrete wavelet transform. The stimulus artifact was due to the pacemaker stimulation voltage of the pacemaker and had a higher slope than the ECG components. The slope was calculated by comparing the difference between two consecutive samples. After proper thresholding of the resulting curve, the position of the stimulus was located. The artifacts were removed from the signal by using spline interpolation to fill the gap as smoothly as possible. The detection of the QRS complex duration was performed with the wavelet transform. The Haar stationary wavelet transform decomposition of ECG signal was selected for its high sensitivity to slope changes in the original signal. The ECG signal was decomposed into the first level approximation (A_1) and details coefficients (D_1). The reconstruction of the signal with setting A_1 to zero provides a new signal, referred to as First Level Details Signal (FLDS). The properties of the produced signals were used for detecting the R peaks, Q and S.

III. Results

The inverse problem of electrocardiography was solved at least 8 times to achieve the optimal results. The best results including the optimal parameters based on the least E_{RMS} between measured and simulated ECG were selected. The initial and optimized values for excitation conduction velocity in different tissues and the location of myocardial infarction are demonstrated in table 1. The initial values are the first estimation of the pathological parameters and were extracted from the previous studies [13]. The 12 standard ECG channels were extracted from the measured 64 channels ECG. These signals are compared to the simulated ECG for 12 standard channels based on optimized parameters of inverse solver before implantation. The simulated and measured Wilson channels ECG pre-implantation are demonstrated in Fig. 3. The signals are highly correlated (up to 0.8) in the interval of QRS complex. Furthermore, the optimal set-up parameters were discovered using the optimization method of CRT (pacing the leads in right ventricular apex (X) and the left ventricular posterolateral area (J) and adjusting the timing delays to 140 ms for A-V delay and 58 ms for V-V delay). The body surface potential maps of the patient models based on the optimal BVP set-up were simulated. The QRS duration of 12 standard ECG channels were extracted in order to estimate the efficacy of the optimization methods for optimal parameters with comparing to the corresponding QRS duration in measured ECGs. The difference of the QRS duration pre and post-implantation for Wilson leads is demonstrated in figure 4. The similar morphology of the QRS duration difference in both measured and simulated cases considering an offset of 15 ms verified the applied optimization method.

TABLE I: Initial and Optimized values for infarction location and excitation conduction velocity in different tissues

Data Set	Initial Values	Optimized Values
Infarction position (x, y, z)	157 132 100	155 102 108
Infarction size (voxels)	20	18.90
ECV in LV myocardium	1065	1315.70
ECV in RV myocardium	1065	1675.40
ECV in RV Purkinje fibers	4862.02	3762.15
E_{RMS}	0.136043	0.123527

ECV: excitation conduction velocity (mm/s), LV: left ventricular, RV: right ventricular, Number of iterations: 272.

IV. Discussion and Conclusion

The clinical practices of University Hospital Mannheim and the simulation results of the BVP optimization are in good agreement. Both methods identified the same left electrode position as being optimal for the therapy. On the other hand, the surgeons would prefer to place the right electrode in the septal position, reasoning that a larger distance between electrodes results in a better coverage of the ventricular area. However, reaching the septal position is an onerous process and to avoid the risk of injuries, usually the right ventricular apex is selected, region, which the simulations proved to be beneficial for the patient. The regions selected as optimal with the computer model, proved to be physically reachable in practice. Simultaneous to the most optimal electrode positions, the computer model also indicates the BVP efficiency specific to all the other electrode positions. Therefore, in case the most optimal configuration cannot be applied in practice because of physical restrictions, the surgeon may choose the second or third optimal electrode configuration, as achieved by the simulations. The timing

delay used in hospital was 100 – 130 ms for A-V delay and 0 ms for V-V delay, while the values obtained in this work are slightly larger. Another relevant advantage of the BVP computer model optimization is that it can be adapted to any individual anatomy and pathology. Additionally, due to the pre-operative advantage, the computer-based method, in comparison to the clinical optimization, leads to a less invasive procedure for the patient and is time efficient, both aspects being beneficial for the patient.

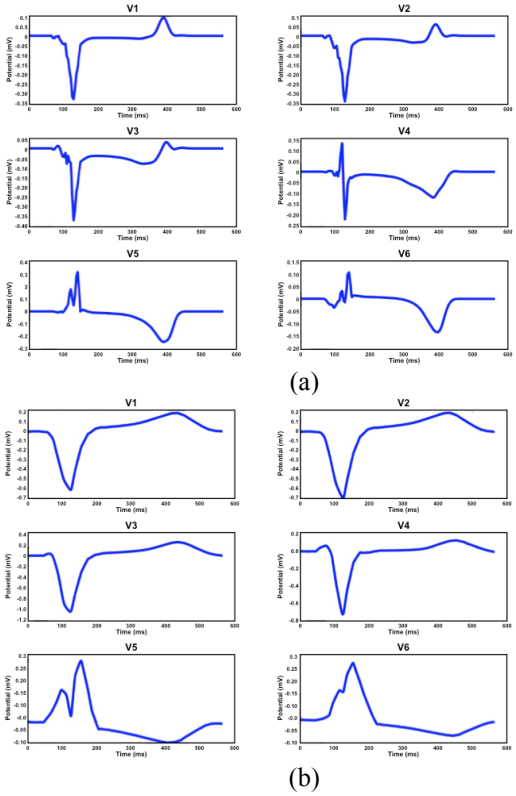

Fig 3: ECG in Wilson leads: (a) Simulated after solving the alternative approach of inverse problem. (b) Measured. (The P-wave in the simulated one does not exist since the atria model was not created due to the lack of information in the MR-data set. The T-wave was not included into the optimization process.)

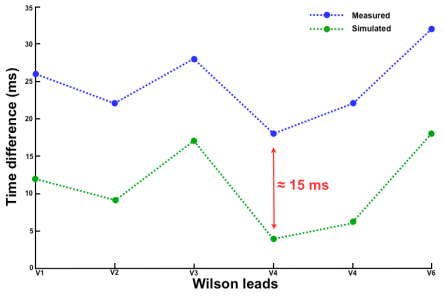

Fig. 4: The QRS duration difference of the Wilson ECG channels before and after implantation.

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