1 Introduction

The study of specialised and relatively well localised conduction tissues of the heart, such as the His-Purkinje System (HPS), has been one of the main objectives of high-resolution magnetocardiography over the last six years. The first magnetic recording of HPS-activity was reported by Farrell et al. in 1978. About a decade earlier Scherlag succeeded in demonstrating HPS activity in man by using an intracardiac catheter technique (Scherlag et al., 1969). In their electrical recordings a spike appeared about 40 ms prior to the onset of the QRS-complex. Four years later Berbari reported that HPS activity could be recorded by surface electrodes using high amplification, proper filtering and averaging techniques (Berbari et al., 1973). The same group also demonstrated that by using a bandwidth of 0.1-300 Hz the PR-segment showed ‘ramp’ patterns which commenced about 40 ms before the onset of QRS and represented HPS-activation.

In 1980 Farrell and his group also demonstrated the ramp pattern by means of high-resolution MCG (Farrell et al., 1980). In the same year Fenici published results that showed a similar ramp structure with certain ‘bumps’ in the PR-segment (Fenici et al., 1980). Both groups, measuring in an unshielded environment, considered the ramps to be due to continuous activation of the HPS. In later work done in the Berlin Magnetically Shielded Room, the high-resolution magnetic data did not always show a clear division of two regions with positive and negative ramps, as reported in the papers mentioned above. In only 10 out of 32 cases single definite ‘bumps’ superimposed on the ramps, which could be related to the His-bundle activation, were found (Erné et al., 1984b). It could be argued that looking only at single high-resolution MCG or ECG time recordings and interpreting certain ‘bumps’ or ‘ramps’ as representing HPS activity is a somewhat doubtful technique. It is difficult to decide which bump or which ramp would be HPS-activity because the PR-intervals show many wavelets and there are no objective criteria to pick out the ‘right’ ones.

A better method of interpreting the HR-MCG data might be to combine all 42 PR-segment tracings into a sequence of spatial maps. Only when the magnetic field distribution shows meaningful dipolar patterns in the sense that associated current dipoles have a location and direction which can be expected from physiological knowledge and simulations, can one interpret this magnetic field as being caused by HPS depolarisation. Simulation studies of the HPS activation do show dipolar patterns in the magnetic field distribution (Erné et al., 1985a).
There remains one difficulty to overcome. Some method has to be found to reliably separate the signal due to the atrial repolarisation from that produced by the HPS activation, as both signals overlap each other in time. Several atrial repolarisation 'subtracting' techniques have been suggested. These are

(a) subtracting the magnetic field pattern found just after the P-wave offset from the signals obtained during the last part of the PR-segment (Ernö et al., 1984b; Fenici et al., 1985)

(b) removing the atrial repolarisation by cross-correlating the static repolarisation pattern with maps of the HPS activation as has been done in high-resolution ECG (Horan et al., 1982)

(c) separating repolarisation from HPS activity by an expansion technique using MCG-spatial eigenfunctions representing the atrial repolarisation pattern (Macaulay et al., 1985).

Lately it has been proved that the last and the first method give similar results (Lamothé et al., 1987), which favours our more straightforward method of subtracting a mean atrial repolarisation pattern.

The aim of our research was to address in a nonshielded environment the following question: Is it possible by means of mapping the magnetic field over the entire chest to distinguish between atrial depolarisation, atrial repolarisation, HPS activity and very early septal depolarisation?

2 Methods

2.1 SQUID-magnetometer

The high-resolution MCGs were recorded with a single-channel magnetometer consisting of a BTi RF-SQUID coupled to a homemade second-order gradiometer. The gradiometer had a baseline of 4 cm and a diameter of 3 cm and was balanced with a three-dimensional system of adjustable superconducting vanes. SQUID and gradiometer were kept superconductive in a 51 fibreglass dewar containing liquid helium. The recordings were performed in an unshielded room in a small brick building on the university campus during daytime. Before digitising the MCG-signal it was filtered by an active notch filter with zero frequency of 100 Hz.

After this procedure the noise level was approximately 20–35 FT Hz$^{-1/2}$ (FT = femtotesla = $10^{-15}$ tesla). To avoid signal distortion no high-pass filters were used.

The magnetic field component normal to a plane that is tangential to the subject's chest has been recorded sequentially at 42 points forming a rectangular measuring grid. The 42 sites with grid space of 5 cm in both directions resulted in a total grid dimension of $25 \times 30$ cm (Fig. 1). Rows are labelled A–G and columns 1–6. The subject was in the supine position during recording, with the measuring plane (i.e. the lowest coil of the gradiometer) as close to the subject's chest as possible, typically 15–20 mm from the chest. Fields coming out of the chest were defined as negative.

2.2 Data acquisition and processing

A 12-bit A/D converter was used to digitise the data at a rate of 1000 Hz. A finite impulse-response digital filter, with a cutoff frequency of 100 Hz and a notch at 50 Hz, was applied to remove the remaining line interference frequencies. A PDP-11 73+ microcomputer system performed real-time averaging of heart beats centred in a 700 ms long window. As a trigger for aligning the MCG, the analogue so-called 'dual-level-cross' QRS-detector was used. At a preset time delay after a QRS-peak has occurred the QRS detector sends a TTL pulse to the PDP-11. Peak detection is based on a level criterium for both ascending and descending slopes of the QRS-complex in the standard ECG-II limb lead, which was bandpass filtered between 0.1 and 30 Hz. This method of aligning the MCG complexes is not affected by drift in the ECG reference lead. The trigger jitter of this hardware QRS-detector was about 0.2 ms (Ros, 1976). The real-time averaging process of the MCG beats was made visible on a video display and served as a useful tool to control continuously the noise level present in the MCG signal and to reject too noisy beats. Only 25 cardiac cycles were averaged at each grid point, which resulted in a very reasonable total recording time of about 60 minutes for each subject, including preparation for proper triggering on the ECG limb lead. When thought necessary the averaged data could be filtered offline with software filters.

To produce smooth isocontour maps at a certain time instant, the magnetic field was computed in a fine regular grid of $20 \times 20$ points from the 42 measured data, using a fifth-degree bivariate interpolation function (Akima, 1978). From this fine grid of 400 interpolated function values isocontour lines were drawn by linear interpolation (Snedeker, 1978). Maps could be made every millisecond from 250 ms before QRS-onset to 450 ms after. All time instants mentioned in this article are related to the onset of the QRS-complex. The QRS onset was taken as the earliest ventricular depolarisation visible in any of the 42 high-resolution magnetic recordings.

Numerical calculations, based upon a volume conductor model which consisted of a realistically shaped torso with four compartments of different conductivity, representing the two lungs and the two intercavitary blood masses, showed that the general features of an isofield contour map were not much influenced by the volume conductor (Peters et al., 1982). The contour indicating zero measured field almost coincided with the one calculated for a current dipole situated in an infinite homogeneous conductor. Although recent model studies report a change of the apparent dipolar source in the presence of intraventricular blood masses, the contributions are small enough to qualitatively interpret magnetocardiographic maps with the simplified infinite-volume-conductor model (Horacek et al., 1986). Therefore by displaying in isofield contour maps