New sensor based on fibre optics for measurement of heart movement

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Abstract—*Innovative fibre-optic sensor technology for measuring the movement of the myocardial walls, and from this the heart chamber volumes, was developed. An optical fibre, with a mirror at its end, is inserted into a catheter located in the heart. An opto-electrical control unit positioned outside the heart contains both the light source and the signal receiver. It generates and couples the light into the fibre and transforms and analyses the reflected signal. With such a system, the movement of the cardiac wall can be continuously measured during each cycle, because the fibre moves synchronously with the heart, and this movement bends the fibre, changing the optical attenuation. Experiments where the fibres were wound around metal cylinders of different diameters revealed a maximum sensitivity of 4% mm^{-1} diameter. The noise signal corresponded to about 1% of the diameter. First tests in a working pig heart showed a high correspondence of the fibre signal with cardiac parameters. Although these tests are promising, further long-term, extensive experiments in preclinical test devices, and later in clinical tests, must be carried out before the new sensor is used in clinical practice. The fibre-optic technique could be used in monitoring devices, assist devices, pacemaker systems or cardioverter defibrillators.*

Keywords—*Fibre optic, Haemodynamic sensor, Cardiac contraction, Sensor-controlled pacemaker, Defibrillator*

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1 Introduction

IN MEDICAL practice, various sensors and measuring devices to record haemodynamic parameters are in use (GEDDES and BAKER, 1989). For intracardial pressure measurements, for example, pressure sensors are located directly within the heart. As deposits from the blood can reduce sensitivity, such sensors will be problematic for long-term use within the heart, until suitable membrane materials become available. Other methods with similar problems allow the measurement of the speed of cardiac contraction, e.g. by using an acceleration sensor fixed at the myocardium. Recently, the cardiokymograph system, a non-invasive method to measure ventricular wall motion, has been revived (GE *et al.*, 1998).

In this paper, a new method for recording the movements of the myocardial wall within the heart is presented; it uses fibre-optic technology, avoiding many problems of interference both with blood and with external electromagnetic fields. This technology has been proposed for several technical purposes and, in medicine, for purposes such as respiratory monitoring (DAVIS *et al.*, 1997; MIGNANI and BALDINI, 1997). To our knowledge, it has not been developed and used for measurements within the heart. On the other hand, there is a strong need for intracardiac measurements for the purposes of heart monitoring, sensor-controlled pacing, automatic defibrillation and control of heart-assist devices.

2 New method for the measurement of myocardial movement

Fig. 1 shows the position of the leads of a pacemaker by superimposing a video sequence of an X-ray scan during one heart cycle. The lower lead is in the right ventricle, and the upper lead is in the right atrium. In the beating heart, the leads move synchronously with the contracting myocardium, producing flexions of the leads.

When an optical fibre is inserted into the lead, it also moves synchronously with the heart. The changes in the flexion cause changes in the optical attenuation of light transmitted. This produces a signal that should correlate with the movements of the heart (HOELAND *et al.*, 1999). The information thus gained can be incorporated in various cardio-technical instruments. In cardiac pacemakers, information about the state of the heart and the control of the autonomic nervous system is required to generate an adequate pacing rate (WERNER *et al.*, 1998; 1999). In implantable cardioverter/defibrillators (ICD), additional information would be advantageous to provide a more reliable criterion for shock application. In this case in particular, such information should be derived from a mechanical signal independent of the electrical signals of the heart, as unequivocal analysis in the case of ventricular fibrillation is difficult. Heart support systems, such as left ventricular assist devices, need additional mechanical information to recognise whether a change in ventricular performance is achieved by the therapy.

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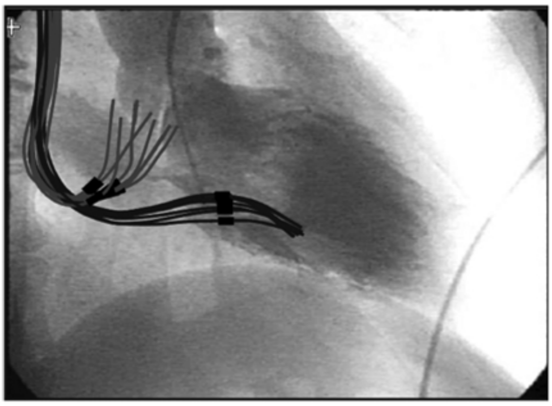

Fig. 1 *Sites of atrial and ventricular pacemaker leads during one heart cycle (7 X-ray images of a video superimposed)*

3 Optical fibres: structure and principle of light transfer

Optical fibres consist of a cylindrical core, with a refractive index n_1, and a surrounding cladding, with a refractive index n_2, smaller than n_1. Further layers (coating), mainly plastic material, protect the fibres against environmental influences and breakage. The difference in the refractive indices can be either stepwise ('step index fibre'), or graded ('graded index fibre'). The different refractive profiles produce different properties in the light transfer (dispersion of 'modes' (GOWAR, 1984)).

Light wave transmission in optical fibres is based on reflection of the light at the interface of core and cladding. If the angle between the light beam and the interface is smaller than the critical angle $\gamma_c = \arccos\, n_2/n_1$, light is totally reflected at the surface of the core and transmitted through the fibre by repeated reflections. Bending of the fibre changes the angle γ, and light can escape from the fibre if the angle γ is larger than the critical angle γ_c (BERTHOLD, 1995; LAGAKOS *et al.*, 1987).

4 Structure of fibre-optic measuring instrument

An instrument for the measurement of the light attenuation of an optical fibre was constructed and optimised for measurements within the heart. Only the end of the fibre outside the heart can be used for coupling the light into and out of the fibre. Therefore the end within the heart was provided with a mirror to reflect the light. Return of the light beam via a second parallel fibre was not possible because of lack of space. Furthermore, a flexion at the end, with a diameter of about 1 mm to return the light, is not attainable at present (LAGAKOS *et al.*, 1987; DAVIS *et al.*, 1997).

Fig. 2 shows the structure of the measurement device. Within the opto-electrical unit, the optical components are indicated by the dotted lines. A beam splitter is responsible for the coupling of the light into and out of the fibre (optical coupler with 2 $\times$ 2 connections, 50%:50% power distribution). The light is transferred from the emitter to the coupler, and one-half of the light power is directed to the reference receiver. The output signal of this receiver serves to stabilise the light source. The other half of the light is fed into the fibre, through which it enters the heart and is attenuated by the flexions.

The light is reflected from the mirrored end and passes back through the fibre to the coupler, where it is divided once again: one part returns to the emitter where it is absorbed or reflected. (As the measurements are quasi-static, the reflections do not cause interference.) The other part reaches the receiver, containing the relevant information about the attenuation of the light. An infrared light diode* was chosen as the light

*HFE4020, Honeywell

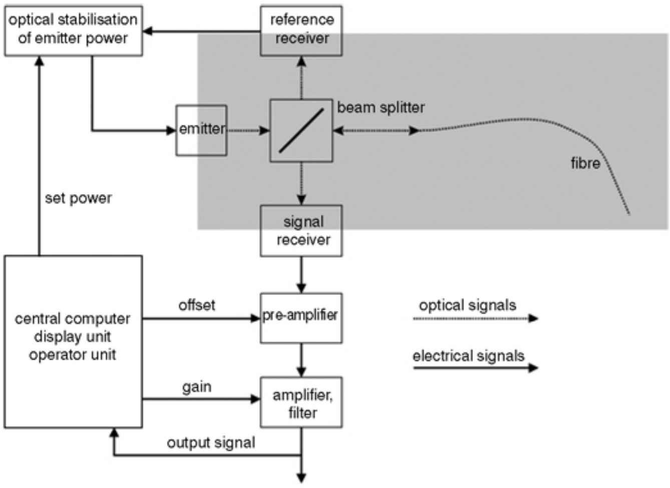

Fig. 2 *Schematic diagram of opto-electrical unit and sensing fibre*

emitter. A non-coherent broad spectrum of light is necessary, because coherent light could disturb the measured signal by interference phenomena (BERTHOLD, 1995). The mean wavelength of the produced light is 850 nm. In the receiver, the optical signal is transformed to an electrical one: the current of a photodiode.

A silicon photodiode† is used as receiver. It works in the short-circuit mode to reduce noise, as the greater capacity in this mode it not harmful in the frequency range used here. As the change in photocurrent lies in the range of a few parts per million, the offset current has to be subtracted from the photocurrent by a current compensation (Fig. 3*a*). The offset current is adjusted so that the mean value of the alternating component is zero. This alternating current is converted via a current-to-voltage converter, the conversion impedance being 10 MΩ. Two amplifiers with programmable gain factors are available to adjust the necessary sensitivity over a wide dynamic range. The amplification factors are displayed and transmitted via a serial interface, so that documentation of the amplification settings is ensured. In a further step, the measured signal is limited to a maximum amplitude of ± 5 V and filtered by a low-pass filter, with a cutoff frequency of 30 Hz.

The power of the optical radiation has to be stabilised (Fig. 3*b*), as the emitter produces fluctuations in the power of the emitted light, depending on temperature (about -0.019 dB per °C) and on time. Therefore the signal from the reference

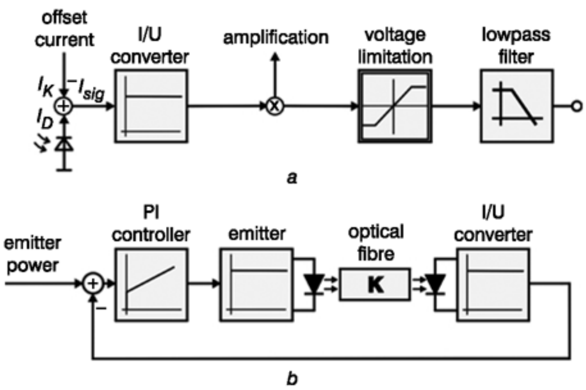

Fig. 3 *(a) Diagram of receiver circuit. (b) Power stabilisation circuit of emitter diode*

†HFD3033, Honeywell

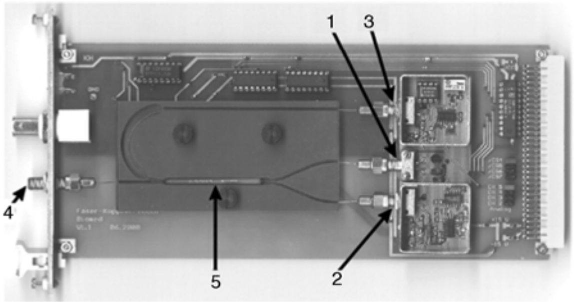

Fig. 4 *Extracorporeal opto-electrical unit. 1: emitter; 2: signal receiver; 3: reference receiver; 4: connector for fibre; 5: coupler/beam splitter*

receiver is amplified and used in a closed-loop circuit to control the power of the optical output.

The calculation and suppression of the offset current, as well as the adjustment of the amplification, is carried out by a microcontroller‡, which simultaneously displays the settings and the measured signals on an LCD (128×64 pixels). The serial RS232 interface of the microcontroller provides the transfer of parameter settings.

The instrument (Fig. 4) is installed in a 19" rack, together with the power supply, the microcontroller unit with LCD and the keyboard, as well as three measurement cards with the electronic components.

5 Preparation and technical testing of measurement fibres

An optical fibre must fulfil several requirements to be suitable as a sensor for the measurement of geometric parameters of the heart.

(*a*) The outer diameter of the coating should not be larger than 250 µm, because the fibre is to be integrated in a pacemaker lead or catheter without any problems. A pacemaker lead usually contains a guide wire to insert it into the heart. This is removed after positioning, and the fibre can be introduced in its place. The guiding wire usually has a diameter of 300–400 µm, and optical fibres with the required diameter are available as standard.

(*b*) The coating must be a biocompatible material, suitable for clinical use. Furthermore, it has to be flexible and mechanically durable, with adequate long-term stability. Biocompatible coatings are available and in use in medical technology.

(*c*) The fibre must have sufficient sensitivity to measure light attenuation.

Based on these criteria (biocompatibility in this step is not being taken into account), 11 fibres were investigated. The relevant parameters of the fibres are listed in Table 1. In the right-hand column, it is noted whether the fibre is suitable for this application.

The mirror at the end of the fibre was prepared as follows: The terminal end was polished in two steps, using polishing foils with a grain diameter of 5.0 µm and 0.3 µm. The quality of the polishing was examined after each step by microscope. Gold is highly suitable as a mirror, because it is chemically stable and, at a wavelength of 850 nm, it has a reflection factor of 96%.

To test the sensing properties of the device, we used a technical setup where the fibres were wound once around metal cylinders of different diameters. The change in photo current was recorded for each diameter (Fig. 5). The fitted

‡80C552, Philips

Table 1 Types of fibre examined

Company	Type	Index	Core, µm	Cladding, µm	Coating/buffer, µm	Aperture	Refractory index core	Refractory index cladding	Coating	Suitability
SpecTran	HCL-M0100T	step	100	125	250	0.22	1.453	1.436	tefzel	OK
SpecTran	HCL-M0200T	step	200	240	375	0.22	1.453	1.436	tefzel	too thick
SpecTran	HCL-M0365T	step	365	400	730	0.22	1.453	1.436	tefzel	too thick
SpecTran	HCL-M0550T	step	550	600	750	0.22	1.453	1.436	tefzel	too thick
Schott	QQ 50/200	step	50	200	280	0.1			acrylate	not flexible
Wave Optics	WT-8MA-532-00F-3.5	gradient	100	140	250	0.29				OK
SpecTran	TCL-MB105H	step	105	125	155	0.22	1.453	1.436	PYROCOAT	OK
Ceram Optec	WF 50/125A	step	50	125		0.22				OK
Ceram Optec	UV 50/125A	step	50	125		0.13				OK
SpecTran	SMC-A0980B	monomode	6.2	125	245	0.11	cutoff:	980 nm	UV acrylate	not OK
Wave Optics	WF-531	gradient	100	140	250	0.29			acrylate	OK

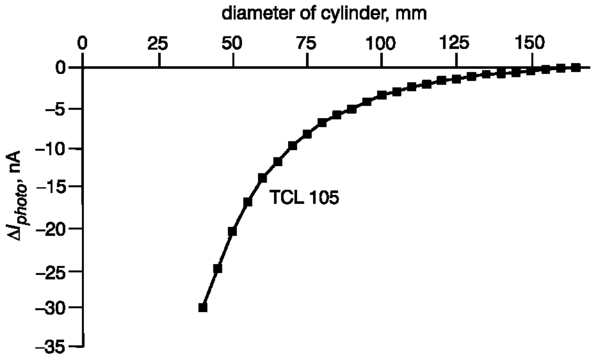

Fig. 5 *Change in photo current as function of bending diameter for TCL 105 fibre*

exponential course is congruent with theoretical considerations. From the various fibres tested (fibres with stepwise changing or linearly changing refraction indices, with outer diameters between 155 and 250 μm, see Table 1), the model TCL-105 from SpecTran, with diameters of 105, 125 and 155 μm for core, cladding and coating was particularly suitable. The values of the change in the photo current in Fig. 5 refer to the value measured at a bending diameter of 165 mm. It must be taken into account that the optical attenuation is also dependent on the arc length wound around the bending cylinder. Greater arc lengths deliver greater damping. The analysis of the parameters, bending diameters and arc lengths obtained adequate results for the values to be expected within the heart. Maximum sensitivity for measurements with the fibres used so far is estimated to be ~4% per mm diameter. The noise signal in the present device corresponds to about 1% of the diameter to be measured.

6 First test of feasibility of measuring method using a working pig heart

To test the prototype of the new sensor system, we used an experimental setup for a working pig heart (Fig. 6) (KLOPPE *et al.*, 2000*a*). The hearts were taken from adult German farm pigs. After explantation from the animals, the hearts were transferred to the laboratory via the standardised transplant mode used for human hearts. This means they were flushed with cardioplegic solution and stored on ice for hypothermia. After preparation, the reperfusion was carried out with the animal's autologous blood, which was rewarmed to the normal body temperature of 37–38°C.

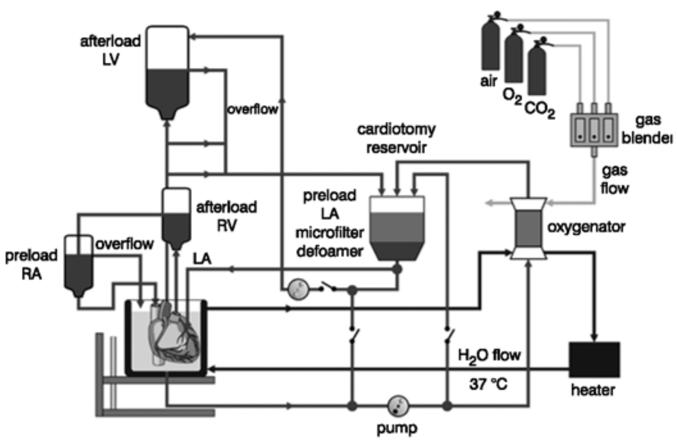

Fig. 6 *Isolated working pig heart. RA = right atrium; LA = left atrium; RV = right ventricle; LV = left ventricle*

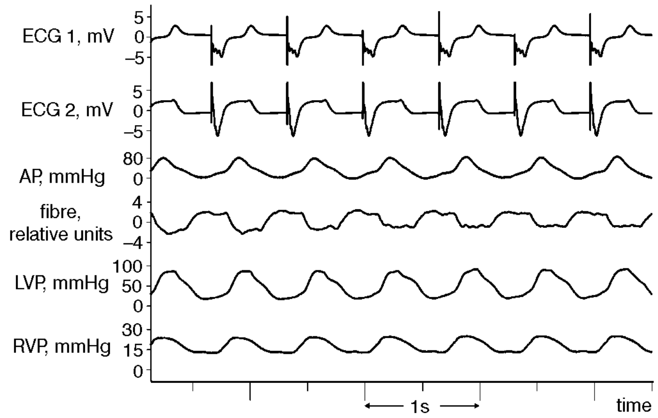

Fig. 7 *Fibre signal during contraction and relaxation of pig heart. Epicardial electrocardiogram and intracardial electrocardiogram: ECG 1 and 2; fibre signal: fibre; aortic pressure: AP; left ventricular pressure: LVP; and right ventricular pressure: RVP*

The oxygenation of the blood and therefore the oxygen supply of the heart were achieved with a standard oxygenator from paediatric heart surgery**. Each chamber of the heart was connected by a specific connector to a reservoir that was filled with blood. These blood storage reservoirs served as air traps and regulatory devices for the control of adjustable pre- and afterload.

With this setup, it was possible to use hearts that have similar physiological performance and haemodynamic behaviour to those of a human heart, to test the new fibre-optic measuring method. To obtain a first confirmation that the new method was biologically feasible, the fibres were inserted into the lumen of ventricular pacemaker leads after the electrode had been positioned at the tip inside the ventricle. As, at present, we are not able to measure continuously ventricular diameters and volumes, cardiac pressures were recorded using tip sensors that were introduced into each chamber of the heart. As the experiment shown in Fig. 7 was based on constant peripheral resistance, a correlation between volumes and pressures can be assumed.

The epicardial ECG (ECG 1), intracardial ECG (ECG 2), fibre-optic signal (fibre), inserted into the right ventricle clinically by the transvenous access, aortic pressure (AP), left ventricular pressure (LVP) and right ventricular pressure (RVP) were recorded. The fibre-optic signal follows the beat-to-beat changes of systolic contraction and diastolic relaxation, as can be seen from the comparison of the fibre and the pressure signals. The fibre signal shows slight superpositions of atrial and valvular movements. If an optimum and constant site is provided for the sensor, the signal is a useful candidate for the measurement of heart movement and of right and left ventricular volume. To monitor left ventricular phenomena clinically using the transvenous access, the fibre has to be inserted into the venous coronary system (KLOPPE *et al.*, 2000*b*). These first *in situ* recordings encourage us to start extensive preclinical and clinical investigations and evaluations, together with our cardiology colleagues.

7 Discussion

The new method has several advantages, compared with other methods of measuring haemodynamic parameters: the sensor is very small and can be integrated in the leads of pacemakers or in catheters without any problems. Because of the extremely small

**COBE Cardiovascular Micro

mass and the high degree of flexibility of the thin fibres, the mechanical characteristics of the leads are not influenced. On the other hand, the leads do not influence the optical fibres. It is advantageous that the sensor does not require any electricity *in situ*. As it consists of a non-metallic material, there are no disturbances to the measuring procedure caused by electrical or magnetic interference.

Further studies of the exact relationship of the signals with the physiological parameters and of the influence of the site of the lead are still necessary, before the method can be used in the clinical environment.

The durability of the fibres is still under scrutiny in an ongoing test. Longevity of at least ten years must be achieved if the method is to be incorporated in implantable instruments, such as cardiac pacemakers or cardioverter/defibrillators. Over this period of time, the heart beats about 370 million times. Until now, the test device operating with threefold frequence has successfully performed about 200 million 'beats'.

To avoid disturbances from movements outside the heart, a special fibre has to be constructed, with low sensitivity outside the heart and high sensitivity within the lead chambers. Therefore two different fibres with different sensitivities have to be connected. Another approach, using the technique of measuring the attenuation in the fibre depending on the local co-ordinate, is rather complex, so that it is unlikely that this latter method will be suitable for implantable instruments.

There are two further important parameters relevant to implantable instruments: the amount of electrical energy consumption and the physical size of the measuring device. At present, the largest power consumer in the measurement setup is the emitter diode. The IR-LED consumes 100 mW. This power can be reduced on the one hand by looking for an emitter with higher efficiency, and, on the other hand, a significant reduction is possible by use of a pulsed rather than a continuous light source.

The geometrical size can be drastically reduced by use of integrated optical components, so that an application in implantable instruments will become possible.

8 Conclusions

The sensitivity of the measurement setup using optical fibres is suitable to measure the movements of the cardiac volume with sufficient signal-to-noise ratio and precision. Evaluation of the first results shows the desired beat-to-beat correlation between the fibre-optic signal and the movement of the heart. No particular technical problems are expected regarding the implementation of the method in extracorporeal instruments. In the case of intracorporeal instruments, further efforts, e.g. with regard to reductions in energy consumption and size, are necessary. These improvements, however, are technically achievable. The precise relationship between fibre-optic signals and the relevant cardiac parameters has to be investigated in extensive preclinical and clinical experiments, particularly with respect to the site of the fibre within the heart.

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