INDUCTIVE TRANSCUTANEOUS COUPLING OF EMG SIGNALS

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The development of the pick-up device for myoelectric signals, reported here, has been carried out as a team work between the Orthopedic University Clinic in Frankfurt/Main and the Electronic Instrumentation of the European Atomic Energy Community Research Estabilishment.

For rehabilitation purposes (e. g. control of artificial limbs), long time operation is required. A variety of solutions has been considered in the past years for the pick-up of myo-electric signals. The three tracks followed are:

— Surface electrodes
— Percutaneous conductors
— Active and passive wireless transmission from implant devices.

The advantages and the difficulties in connection with long time application of surface electrodes and percutaneous conductors are well known and shall not be discussed here. The active, implanted transmitting devices are technically very effective, but suffer from the requirement of a real long time energy source of very small size. Reachargeability through non galvanic coupling during sleep periods for instance is a certain but not very comfortable way out. The passive implanted transmitting devices are therefore a logic answer to the problems. All of them "live" from energy radiated somehow into the body. This energy is then converted, modulated and retransmitted (e. g. Kaiser, Kaiser Lab., Copenhagen) or reflected respectively absorbed under modulated conditions (e. g. EMGOR, Reilly, West Hendon Hospital, London).

Since the implanted devices and the percutaneous conductors have in common the important advantage, compared with surface electrodes, of a better selection of the desired muscle signal because of the reduction of crosstalk, we felt justified, in spite of the existing solutions with good performance, to try still another method using implanted transmitters. We think that our solution, because of its technical simplicity and therefore, as we hope, good long time reliability might be under certain circumstances a useful contribution to the pick-up problem.

Our Magnetic Core Antenna does not require an implanted energy source nor the supply by radiated or somehow transferred energy since it is converting the myoelectric energy directly into an alternating magnetic field. Therefore it belongs to the group: Active wireless transmission devices, however, apart from the usual ones, without any power supply provision (because the energy of the myoelectric signals itself is utilized to drive the coil, being in other words one of the two coils of a transformer).

The biotechnical parameters given were rather discouraging in the beginning and somewhat contradictory:

1) Very low transmission energy available (in the order of a pW) and rather strong parasitic signals from the ambient, both calling for a long magnetic dipole, hence great dimensions.

2) The myo-electric energy, being concentrated at low frequencies (below 500 Hz), and high source impedance (e. g. 5 kΩ), both require matching a high inductivity by an open magnetic circuit, which in turn means great dimensions.

3) Compatibility for subcutaneous long time implantation, also where not very much space is available, requires that the size be small.

It is evident that we had to compromise in some respect. The only rather free parameter has been the transmission distance, since we wanted to avoid percutaneous conductors but not necessarily intracorporeal leads. Therefore the transmission antenna coil has to be fixed subcutaneously and in some cases eventually within the fatty tissue, whereas the electrodes are placed inside or at the muscle fascia and are connected to the coil by two insulated intracorporeal leads. The extracorporeal pickup coil will then be fastened on the skin near and with the axis parallel to the transmission coil (clear distance between the two coils e. g. 5mm). Several geometrical forms of the magnetic core have been considered. It resulted that the stretched cylindrical coil seems to be most effective and less sensitive to small displacements with respect to the extracorporeal pick-up coil. We constructed, finally coils of 3mm diameter of encapsulation and 20mm length, after experiments to obtain the maximum number of ampere-turns for a given internal source resistance, a given cource voltage and frequency, and finally coil dimension. These conditions lead to a rather complex calculation considering also the optimum ratio between magnetic core and winding cross section and the changing space factor for different wire diameters. For a measured source resistance of 7 kΩ (with small electrodes) and a core material of high permeability, 9000 turns of 0.025 mm enameled wire on a core with a cross-section of 1.2×1.2 mm seemed to be a good compromise between the theoretical optimum values and still tolerable production difficulties. The achieved inductivity has been 900 mH; the active resistance about 2 kΩ.

The core was first varnished with Dow Corning Sylgard. During the winding the wire has been wetted with Sylgard and finally a Silastic coating was given to the coil after fixation of the electrode wires by two threads for surgical suture and medical grade glue around the coil. The threads are fixed in their center. Thus the 2×2 free ends are available for suturing the coil to the tissue. Other types of the coil have been made differently: the core has been isolated with Urethane Sealcoat spray (CRC), the electrodes fixed to the coil with Araldite D with hardener HY956 (CIBA); finally the coil has been covered with Sylgard. The core ends have been smoothed by two caps of Silastic.

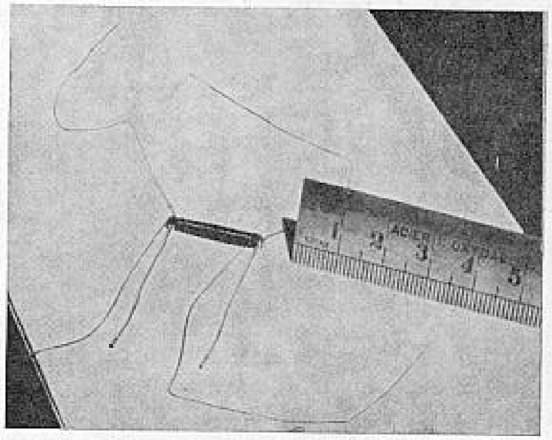

Fig. 1. Implantable coil, coated with Sylgard, with electrodes, and Catgut threads for fixation.

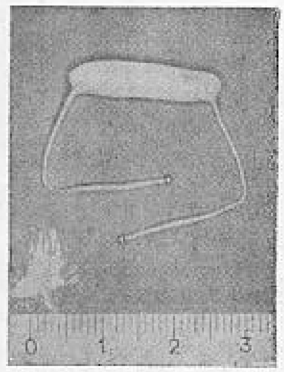

Fig. 2. Implantable coil, coated with Silastic.

The electrode wires have not been made of Karma-wire or equivalent because of the losses in that type of wire (e. g. 2×5cm→220Ω). On the other hand the electrode wires are exposed to mechanical movements. Thus to avoid breakage, chemically very pure gold wire of 0.4mm diameter. giving a high ductility, has been used. These wires were insulated by Silastic tubes. For most of the in vivo tests the electrodes have been gold spheres of 1−1.5 mm diameter with highly polished surfaces (Figs. 1 and 2).

Several antenna coils have been implanted into the musculus quadriceps femoris of a dog. After cutting the fascia, the electrode spheres were inserted between the muscle fibres in distances from 1.5 to 2.5 cm measured along the fibres. The muscle fascia was sutured and the electrode wires in some cases fixed at the fascia by the same suture. The coil has been fixed first on the fascia, later on the inner skin surface. Implantations of the electrodes in man with extracorporeal coils have also been made. For the extracorporeal pick-up coil another similarly stretched cylindrical coil of 25 mm length has been used (Fig. 3).

The output signal from the pick-up coil is processed in an amplifier chain (Fig. 4) starting with a preamplifier consisting of a pair of low noise transistors and an integrated differential amplifier, followed by a T-notch filter for 50 Hz, a band-pass amplifier, a low--pass filter to increase cut-off of higher parasitic frequencies and an output buffer. This amplifier is a laboratory development for tests. An amplifier with low energy consumption is under development.

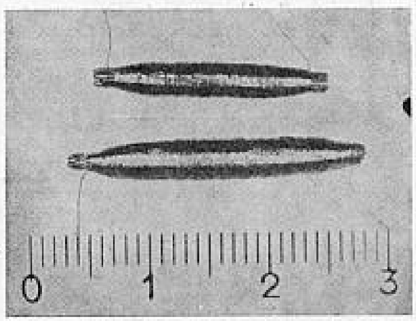

Fig. 3. Implantable coil uncoated (above) and extracorporeal pick-up coil (below).

The results of the in vivo tests are rather encouraging. For medium muscle contraction and 4 mm clear distance through the skin the signal to noise ratio has been about 30 dB using a second extracorporeal coil for compensation of parasitics. The input noise

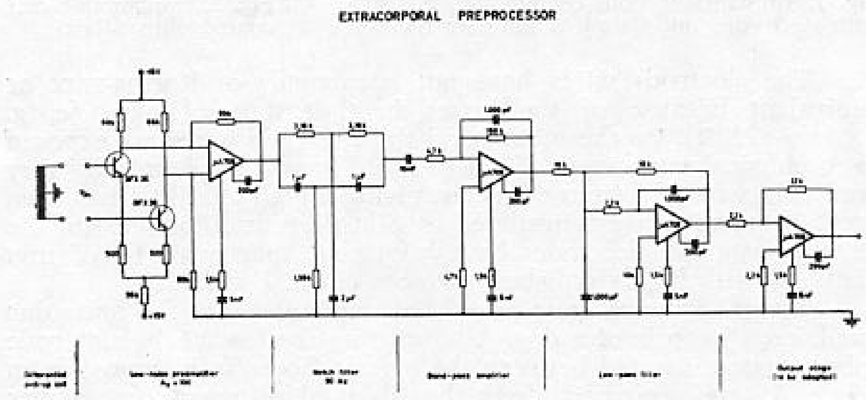

Fig. 4. Schematic circuit of extracorporeal amplifier chain.

level of the extracorporeal device, allows for more reduction still of parasitic signals. At present we have at the extracorporeal pick-up coil a signal voltage for medium contraction of about 160 μV pp. That means at the implanted electrodes, and hence at the intracorporeal transmission coil, 1.120 mV pp. (Fig. 5; time scale 10 ms/cm; vertical scale, amplification considered, about 0.5 mV/cm).

Nevertheless we feel that for long term operation where performance might drop due to electrode encapsulation by growing tissue, and for application with weak source signals the results should be improved further. The easiest way should be to decrease the source resistance by increasing the electrode surface. With this method the energy delivered to a matched transmission coil will increase because of an increase in current availability and because of an increased pile-up effect.

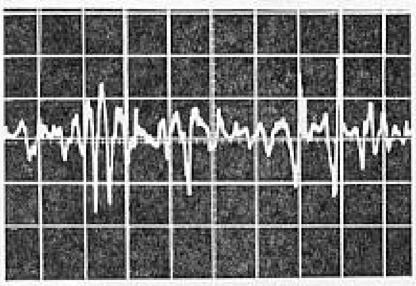

Fig. 5. Obtained EMG-signal at output of extracorporeal amplifier.

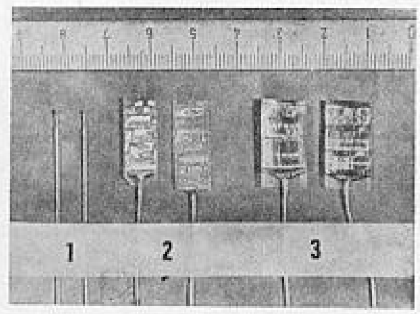

Fig. 6. Small spherical electrodes (1) and laboratory models of large surface electrodes (2 and 3).

The first prototypes of implantable electrodes with considerably larger surface to be placed on the inside surface of the muscle fascia or between muscle fibres are shown in Figure 6. One hundred and fifty thin gold wires (20 μ diameter each) stranded together, form the electrode leads, and are placed side by side on a Silastic sheeted Dacron texture and fixed with 3 layers of medical grade glue. The area of the surface is about 150 mm². With this area the source resistance could be expected to be below 500 Ω. In such a case the antenna coil could be given a much lower impedance, thus energy transfer is increased. Furthermore, the larger diameter coil wire, should give still higher reliability. In vivo tests with these new electrodes and changed coils have just been started.

As a conclusion it may be stated that magnetic coupling of the original myo-electric energy itself is possible for those cases where the transmission coil could be fixed subcutaneously, and promises to be useful and reliable for long-term application because of the simplicity of the implanted components. It is felt that, if the fixation of the extracorporeal pick-up coil could be achieved in a rather stable position relative to the transmission coil, a good proportional and reproducable signal according to the amount of muscle contraction could be gained. Due to the geometrical selectivity of the pick-up electrodes it is believed that this method could be applied also to provide a simultaneous read-out from different muscles in the same region (multichannel operation) all feeding their own subcutaneous antenna coil for transmission. Because the low frequency field of parisitic signals is in most practical cases sufficiently homogenous, the extracorporeal compensation coil for parasitic signals could be used to serve all extracorporeal pick-up coils, mounted not too far from each other, thus the number of these pick-up coils which equals the number of read-out channels is increased by only one additional compensation coil. We imagine that the implantation of the transmission coils and the related electrodes might be done preferably, immediately upon amputation.