A METHOD OF CREATING A MUSCULAR PUMP FOR ASSISTED CIRCULATION

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Existing mechanical apparatuses for assisted circulation (AC) have various disadvantages. First, the traumatic action of mechanical pumps on blood has not been eliminated [8], especially during prolonged use [9]. Second, the large size of the necessary leads [1], the power consumed by them (from units to tens of watts), and the connected problem of dissipation of heat inside the body [5], necessitate certain components of AC apparatuses being located outside the body. These disadvantages have prevented the development of completely implantable AC apparatuses.

The writers suggest a new model of blood pump for the most widely used method of AC, namely aortic counterpulsation. This model differs in principle from mechanical pumps in that it is created from the body's own skeletal muscles and it is controlled by electrical stimulation.

The muscular blood pump is shown schematically in Fig. 1. To make the pump in some experiments we used the autologous left pectoralis major muscle, which was denervated, isolated on an arteriovenous pedicle, and transplanted into the left pleural cavity. In other experiments the autologous skeletal adductor longus muscle was used, and was isolated from the right or left thigh on an arteriovenous pedicle, then freely grafted in the left pleural cavity and its vessels anastomosed with the subclavian artery (A) and vein (V) by the apparatus described in [7]. Next, electrodes were introduced into the thickness of the autologous muscle perpendicularly to the muscle fibers and throughout its width, the proximal portion of the descending thoracic aorta was wrapped with the autologous muscle, and its edges were sutured.

A myocardial electrode was sutured into the wall of the left ventricle in an avascular zone. A hook-shaped electrode was used for this purpose in acute experiments. Electrodes from the autologous muscle and the heart were connected to a pacemaker (Fig. 1) and the muscle was stimulated under cardiosynchronized conditions by the R wave with a multiplicity N_R = 1-8. The delay T_d and duration of the stimulating burst T_b were established, just as in aortic counterpulsation [8], depending on the shape of the aortic pressure curve.

An apparatus of our own design [4] was used for stimulation. The principal characteristics of the apparatus are: mode of operation asynchronous or cardiosynchronized, duration of delay of burst of stimulating pulses relative to R wave T_d = 50-500 msec, duration of burst T_b = 50-500 msec, duration and frequency of pulses in burst 0.5 msec and 60 Hz respectively; conversion factor of R waves (multiplicity of synchronization) N_R = 1-8, blocking of extrasystoles provided over whole range of delays T_d, output voltage within the range U = 0-18 V (load capacity up to 60 mA), output current within the range I = 0-12 mA (load capacity up to 8 V), independent power supply from 7D-01 battery.

Altogether 20 acute and three chronic experiments were carried out on mongrel dogs weighing 18-30 kg. Typical curves of synchronized operation of the autologous muscular pump under counterpulsation conditions with multiplicity N_R = 1 are illustrated in Fig. 2. The aortic pressure (BP) was measured in the ascending part of the thoracic aorta. The beginning of stimulation can be seen as the appearance of bursts of stimuli on the ECG (arrows), ele-

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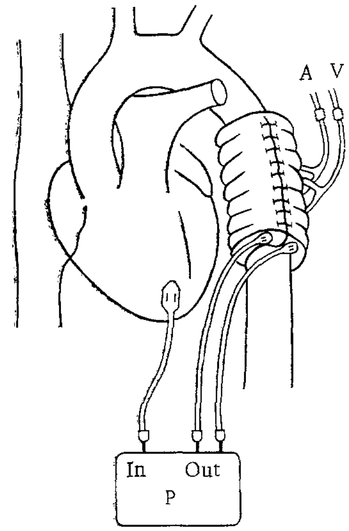

Fig. 1. Diagram of muscular pump for aortic counterpulsation. A) Artery; V) vein; P) pacemaker.

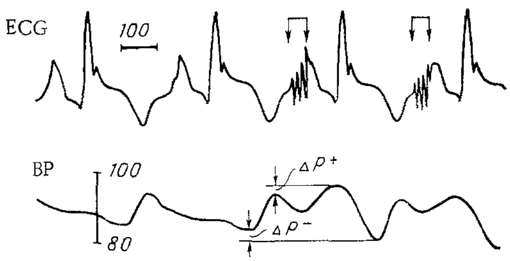

Fig. 2. ECG (lead II) and BP before and during operation of the muscular pump. Cardiac rhythm 167 beats/min, multiplicity of synchronization $N_R = 1$; $T_d = 250$ msec; $T_b = 75$ msec; U = 4 V; scale: vertically, in mm Hg; horizontally, in msec.

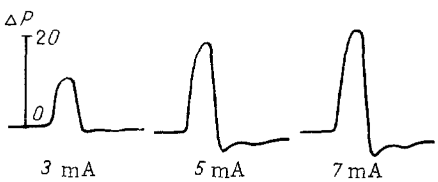

Fig. 3. Gain of pressure ΔP (in mm Hg) created by muscular pump during stimulation of muscle by bursts of pulses of current of different amplitude at $T_b = 200$ msec and repetition frequency of bursts 200/min.

vation of the diastolic BP, and a decrease in the end-diastolic pressure. In this case the diastolic BP was increased to a value $\Delta P^+ = 2.5$ mm Hg above the systolic pressure, and the end-diastolic pressure was reduced by $\Delta P^- = 3$ mm Hg, evidence of the positive effect of the AC [8].

Special experiments were carried out to determine the throughput of the muscular pump. A rubber balloon, connected to the transducer of an electromanometer was wrapped in the adductor longus muscle on an intact arteriovenous pedicle. An excess pressure of 100 mm Hg was created initially in the balloon. Electrodes were then introduced into the muscle and it was stimulated. In the course of stimulation changes of pressure in the balloon ΔP on account of contraction of the muscular pump were recorded. As Fig. 3 shows, using a comparatively weak stimulating current the muscular pump was able to create an excess pressure of $\Delta P > 20$ mm Hg. The power consumed by the pacemaker did not exceed 50-100 μW.

Three chronic experiments with an implanted aortic counterpulsation system were carried out by this method. Implantable asynchronous muscle stimulators were made in the body of an ÉKS-2 pacemaker [3]. After the operation, the working of the implanted pacemaker was monitored by means of a no-contact rhythm indicator [2] and the working of the muscular pump was monitored by recording the surface rheogram of the aorta [6]. Dogs with functioning autologous muscular pumps survived from 3 to 5 days.

By the method suggested above it is thus possible to make a blood pump from autologous skeletal muscle, cardiosynchronized pacing of which ensures effective aortic counterpulsation. It was shown that a completely implantable AC system of small size can be created in principle.

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