From "MEC 93," Proceedings of the 1993 MyoElectric Controls/Powered Prosthetics Symposium
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CONTROL OF POWERED PROSTHETICS USING BEND-ENHANCED FIBRE OPTIC SENSORS

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INTRODUCTION

Powered prosthetics have become the accepted method of replacing limb function lost by traumatic or congenital amputations [1]. The control of these devices is accomplished by either mechanical switches or by switches based on the level of myoelectric activity. Although both approaches provide excellent solutions, there are problems inherent with mechanical and myoelectric control (MEC). Because the mechanical switches used in the prosthetic industry must be small, they lack durability and often fail. For myoelectrically controlled systems a total contact socket is required to minimise the effects of motion artifact and to allow continuous detection of the myoelectric signal (MES). This is often difficult to obtain and leads to signal contamination by 60 Hz interference. Furthermore, perspiration disrupts normal myoelectric signal detection leading to a loss of control. To overcome these problems the Institute of Biomedical Engineering has begun to investigate the use of a *bend-enhanced fibre (BEF) optical sensor* to construct an in-socket transducer for prosthetic control.

OPTICAL FIBRES

Optical fibres are cylindrical structures intended to guide optical signals. They are used routinely in the communications industry to route both telephone and computer signals over great distances. The optical fibres can be made from either plastic or glass, however the glass versions have superior transmission properties. A typical glass fibre used for communication applications is shown in Figure 1. In this diagram, the actual conducting structure comprises the core and the cladding, the remaining structure providing both mechanical and environmental protection.

While there are several types of optical fibres, all depend on the difference in refractive index between the core and the cladding to guide the optical signal. The cladding has a lower refractive index than the core, which leads to propagation via total internal reflection of the signal. Typical sizes of bare fibres (no protective layers) range from 50 µm to 500 µm.

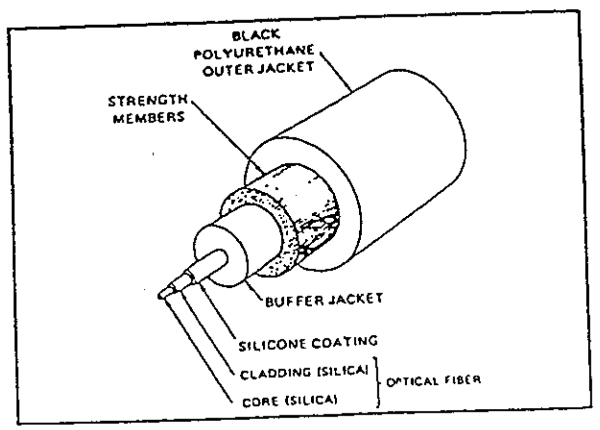

Figure 1: Typical optical fibre structure used by the communications industry.

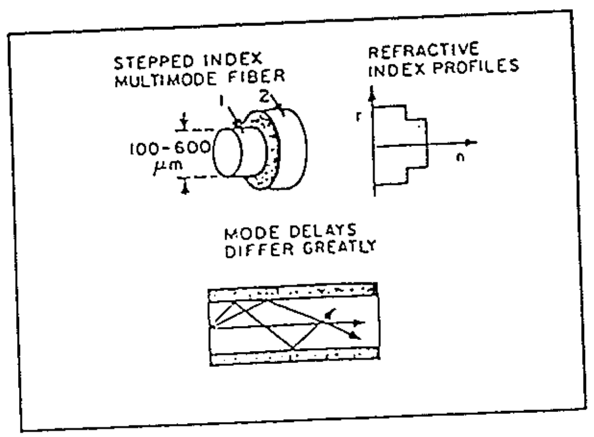

Figure 2: Propagation of signals down an optical fibre by internal reflection.

Recently, applications of optical fibres apart from communications have emerged. The effects of mechanical deformation of the fibre can be used to create a sensing element [2]. If a fibre is bent, small amounts of light are lost through the cladding of the fibre. This increase in transmission loss can be measured and correlated with the mechanical deformation.

BEND-ENHANCED FIBRE OPTIC SENSORS

While all optical fibres exhibit transmission loss under mechanical deformation, the sensitivity of such "microbend" tranducers is very low. Recently, a technique has been developed in which this sensitivity is greatly increased [3]. Bend-enhanced fibre (BEF) sensors are made by treating optical fibres to have an optically absorptive zone along a thin axial strip a few millimeters long as shown in Figure 3. Light transmission through the fibre past this zone then becomes a function of curvature.

The BEFs exhibit a sensitivity three orders of magnitude greater than untreated fibres. Directionality and polarity of curvature are preserved in the transmission function, over a linear range covering 5 orders of magnitude, centered about zero curvature. The simplicity and potential low cost of the sensors and their attached instrumentation makes it likely that they will be used in a wide variety of practical applications where optical fibres are preferable to electrical sensors.

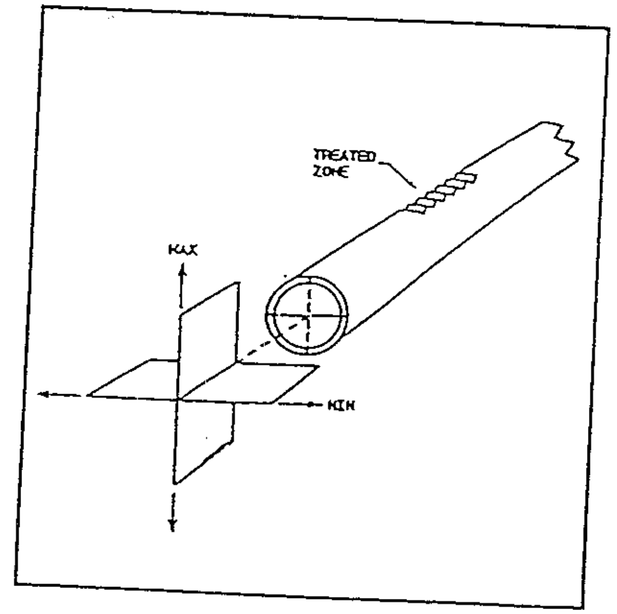

Figure 3: BEF sensor is an optical fibre with a treated strip. Planes of maximum and minimum sensitivity are indicated.

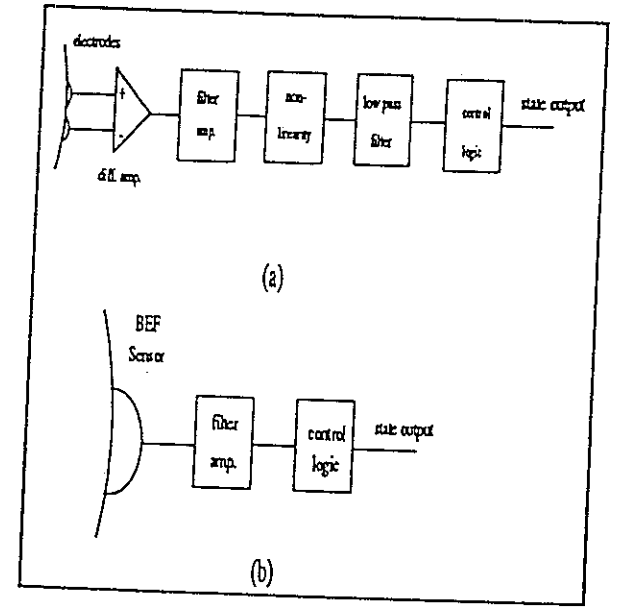

Figure 4: Simplified block diagram of (a) myoelectric control system and (b) BEF control system.

PRELIMINARY INVESTIGATION

At the Institute of Biomedical Engineering a preliminary investigation is under way into methods by which a prosthetic control signal can be derived from pressure changes within the socket. A contraction of the stump musculature causes a small deformation in the socket. By using a BEF mounted within the socket, this deformation can be transduced into a control signal which reflects the pressure exerted by the stump on the socket. In a manner analogous to myoelectric control, a system based on signal amplitude or signal rate of change can then be constructed. A proportional control signal can also be extracted from this transducer to be used to control the speed of the device.

One of the major advantages of using BEFs as sensing elements is in the simplicity of the processing necessary to derive the control signal. In a myoelectric system an electrode pair placed over an active muscle will transduce the underlying ionic current flow in the muscle into an electrical signal. This low amplitude signal is contaminated with 60 Hz interference and must be differentially amplified and processed through a suitable nonlinearity and filter to extract a control signal. This control signal is then compared to a set of switching levels to determine system state.

The BEF sensor, on the other hand, is a complex transducer requiring an LED light source and photodetector. The sensor must be mounted on a structure which exhibits an angular displacement with applied force. Because of the microradian sensitivity of the sensor, the magnitude of the displacement can be very small. Due to the nature of the BEF signal, post processing is simplified, requiring no nonlinear element.

The block diagrams of Figure 4 show the relative complexity of both the proposed BEF control system and a conventional MEC system. A preliminary investigation demonstrated the feasibility of this approach but highlighted a problem inherent with the sensor. Residual pressure against the sensor varied considerably due to small alignment changes within the socket. This baseline drift was stabilized by surrounding the sensor with a foam ring to introduce a pressure dead-zone. The foam must be compressed before the sensor detects a change in pressure.

EXPERIMENTAL PROTOCOL

Once the drift of the transducer had been compensated, a more controlled investigation into the nature of the control signal was initiated. It was thought useful to compare the control signal from a conventional myoelectric setup and that attainable from the BEF transducer. Areas for investigation were the speed of the transducer and its sensitivity.

A plastic cuff was manufactured using conventional prosthetic materials which was used to house the BEF based pressure sensor. This was fastened to the inside surface of the cuff using double sided tape. The cuff was positioned over the bulk of the biceps muscle in a non-amputee volunteer subject and secured in place using a velcro strap. A pair of passive stainless steel electrodes were also located close to the sensor so that a comparison could be made between the myoelectric signal and pressure signal for the same contraction level.

The passive electrodes were connected to a conventional differential amplifier with a gain of 40,000 to provide the raw MES. This signal was further processed using an RMS converter to provide a signal whose level reflects the degree of muscle contraction. This is very similar to the usual method of processing a MES for prosthetic control. Two RMS converters were used in parallel so that a comparison between time constants could be made. The BEF sensor was interfaced to an amplifier with a much reduced gain, x20 with no additional signal processing.

The subject was instructed to perform a series of contractions of the biceps muscle while the signals from the electrodes and BEF sensor were recorded simultaneously. A brief study of more complex muscular activity involving rotation of the forearm was also investigated.

EXPERIMENTAL RESULTS

Experimental results of using a BEF to extract signals suitable for prosthetic control have been very encouraging. In Figure 5 shown below, the output from the sensor is compared with the processed MES obtained for the same sequence of contraction patterns. The graph shows the effect of 50 ms and 200 ms time constants for the MES processing along with the direct signal from the BEF. The correlation between all three traces is immediately obvious. While the 50 ms MES signal responds the quickest, the time constant of the BEF is estimated to be of the order of 100 ms. This compares favorably with typical time constants in commercial myoelectric controls.

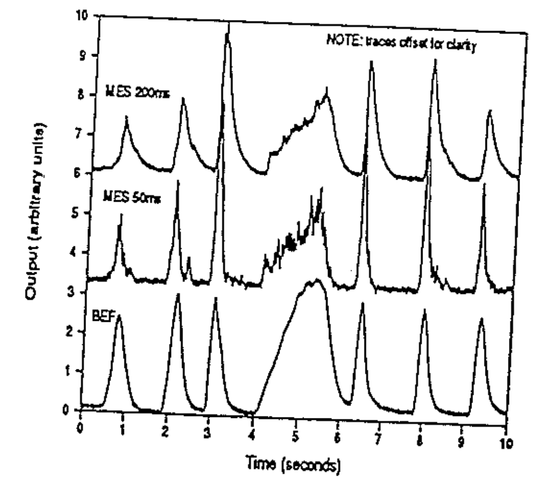

Figure 5: Comparison between BEF signal and processed MES.

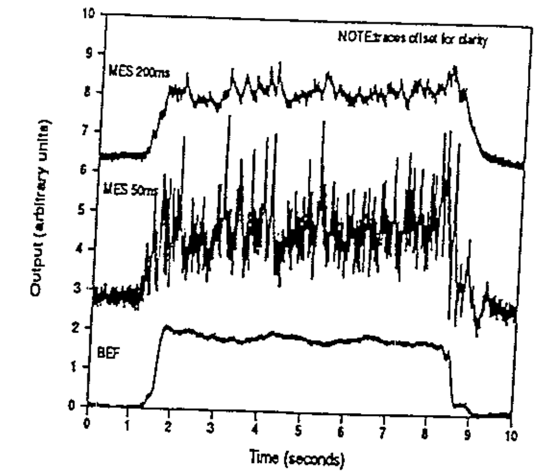

Figure 6: Comparison of steady state varience between MES and BEF signals.

In Figure 6, the response of the BEF to a steady state contraction is illustrated. Comparison to the signal obtained using the MES is also made. It is obvious that the variance in the BEF output is much lower than that of either the 50 ms or 200 ms processed MES, and therefore provides a more reliable signal. If amplitude control is implemented, then the BEF provides a stable signal for amplitude coding beyond the usual limit of three levels, while still maintaining a quick response.

The most novel feature of the new sensor is shown in Figure 7. The output is truely bipolar - responding to pressure changes in both directions. As the BEF is bent in the direction of the treated region, the transmission loss decreases.

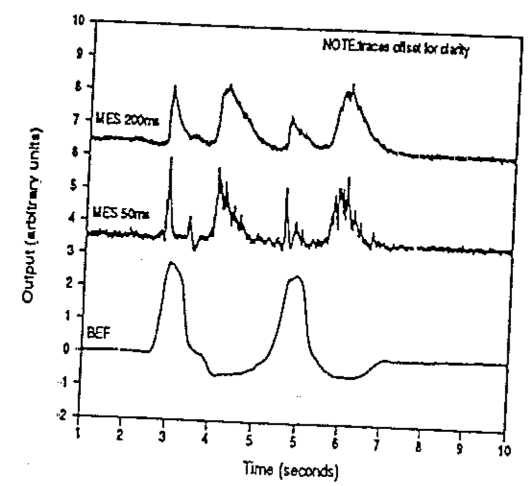

Figure 7: Bipolar nature of the BEF compared to the MES.

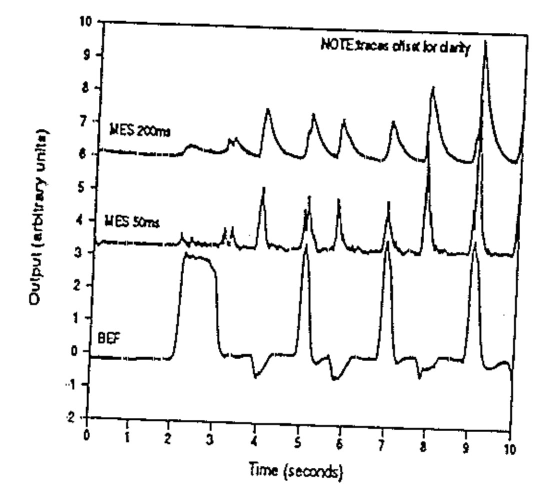

Figure 8: Medial & lateral humeral rotation

The signals recorded in Figure 8 show the effect of medial and lateral rotation of the upper arm. While the processed MES shows distinct muscle activity, discrimination between the two movements cannot be made. However, the bipolar nature of the BEF signal makes this discrimination trivial. Consequently, the use of this sensor for multi-function control is a distinct possibility.

PROSTHETIC APPLICATION & FUTURE WORK

On the basis of these early results a 3-state hand control using a BEF sensor has been implemented. This signal was processed using a circuit similar to that which is presently used in the UNB 3-state myoelectric control system. This level coded system was briefly evaluated using a normal limbed volunteer. Control of the hand with the BEF control system was easily achieved and required a strategy on the part of the user not unlike that used for myoelectric control.

While the introduction of the foam pad into the sensor proved satisfactory for the initial experiments, permanent compression of the ring over time could lead to system malfunction. Consequently, further work has been targeted towards other methods of overcoming the problem of drift with the tranducer. One solution is to implement a system which determines the control signal from the rate of change of output from the sensor. Another alternative is to laminate one or more sensors within the socket wall. This would reduce wear on the sensor during use. Investigation into both these areas is planned for the near future.

CONCLUSIONS

Control systems based on bend-enhanced fibre optic sensors hold promise for prosthetic use. Unlike myoelectric control, they are insensitive to 60 Hz interference. Unlike mechanical switches, they have no moving parts. System malfunction due to perspiration has been a problem for wearers of myoelectric control systems in warm climates. This is eliminated by the new transducer as it does not require total socket contact or contact with the skin surface.

Further work will be done to investigate the optimum sensor mounting and examine alternate control strategies. In addition, the inherent bi-polar operation of the sensor lends itself to the control of more than one function from a single site.

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ACKNOWLEDGMENT

This work was supported by funding from the Canadian Space Agency and the Province of New Brunswick under a cooperative agreement with the Atlantic Canada Opportunities Agency.