Introduction

The electrosurgical unit (ESU) is a useful and necessary tool for surgery. However, when the ESU is activated, the ECG monitor is exposed to extraordinary noise interference generated by it. The ESU poses unique problems (Hujita and Webster, 1974; Yelderman et al., 1983; Uytendaele et al., 1978) because of its large amplitude and wide bandwidth interference, and the mode of interference is dependent on the ESU configuration. Among them, the spark-gap ESU radiates a large magnitude of wide bandwidth interference. The recent type of ESU which consists of vacuum tubes or modern solid state circuits generates a high frequency (HF) output range from 0.5 to 3 MHz which is modulated at a 60 Hz line frequency or its harmonics.

Whatever type of ESU is used, a spark discharge occurs at the tip of the active electrode of the ESU, and a further complication is that the spark discharge modulates the high-frequency output of the ESU. This modulated HF interference is rectified by nonlinearities in the input stage of the ECG amplifier and, as a result, the interference is converted to a low-frequency interference. This modulated HF interference is rectified by nonlinearities in the input stage of the ECG amplifier and, as a result, the interference is converted to a low-frequency interference. Unfortunately, the low-frequency interference has frequency components which are the same as those of the ECG signals and electronic filters can barely separate them from ECG signals. Furthermore, the magnitude of the ESU interference is so large that it saturates the preamplifier of the ECG monitor.

However, the ECG monitor must still be designed to operate in this environment. To solve these problems, the techniques of electrosurgical interference cancellation using low-pass filters (Chu et al., 1985) and signal processing (Yelderman et al., 1983) have been described.

In general, demodulation of the interference in the amplifier can be suppressed by low-pass filtering all inputs of the amplifier, and typically frequency components above 30 Hz are cut off (Uytendaele et al., 1978). As a result, low-frequency interference which cannot be removed remains after low-pass filtering. Furthermore, a slight component imbalance of the low-pass filter converts a common-mode noise signal to a normal-mode (differential) noise signal which can not be eliminated in the differential amplifier. This paper proposes a fibre-optic ECG monitoring instrument which effectively reduces ESU interference and enhances ECG monitoring during surgery without such low-pass filters.

Analysis of the possible interference path

ESU interference occurs in three possible modes. The first is conduction via the AC power line, by which the HF interference is carried via the AC power line into the monitoring device and interferes with the sensitive amplifier circuits in it. The second is radiation, by which the HF interference is transmitted via electric and magnetic field coupling into the cables of the monitoring device and its circuits. The third is conduction via the patient, by which the HF interference is carried via the patient's body into the monitoring device and produces large noise voltages across the electrodes.

Fig. 1 shows a schematic ECG amplifier circuit and the possible ESU interference path combinations. $Z_{11}$ and $Z_{22}$ are the impedances of the two differential input electrodes.
of the ECG, and $Z_{ac}$ is the right leg (RL, common) electrode impedance. $Z_{cm1}$ and $Z_{cm2}$ are the common-mode input impedances of the amplifier. $Z_{g}$ and $Z_{o}$ are the impedances earthing the monitoring device and the ESU, respectively. $I_{o}$ is the current flowing to the earth through the common electrode impedance $Z_{ec}$ and the earth impedance of the monitoring device $Z_{g}$. Consequently it flows back to the ESU through the earth impedance of the ESU ($Z_{b}$). This unwanted earth current $I_{o}$ produces the common-mode noise voltage ($V_{cm} = Z_{ec} \times I_{o}$) across the common electrode impedance $Z_{ec}$.

Several investigations of the interference of powerline frequency on biopotential recordings have been described (HUHTA and WEBSTER, 1974; THAKOR and WEBSTER, 1980; WINTER and WEBSTER, 1983). From this research, we present the following equation for the ESU interference.

Total interference voltage at the amplifier input:

$$V_{n} = \frac{V_{p}}{K} + V_{em} + Z_{b}I_{b} + (Z_{b}I_{b} + Z_{c}I_{c}) \times \left[ \frac{1}{CMRR} + \frac{Z_{cm1}}{Z_{cm1} + Z_{d1}} - \frac{Z_{cm2}}{Z_{cm2} + Z_{d2}} \right]$$

(1)

The first term $V_{p}/K$ represents conducted interference via the AC power line. Here, $V_{p}$ is a noise voltage representing noise coupling into the power line from the ESU, and $K$ is the immunity factor of the monitoring equipment:

$$K = \frac{\text{noise voltage induced in the power line}}{\text{noise voltage appearing in the recording}}$$

(2)

The second term $V_{em}$ represents the interference induced into the cables of the monitoring device and its circuits by the power line. However, this can easily be eliminated by using shielded cables and decreasing the size of the conductive loop (HUHTA and WEBSTER, 1974). $Z_{b} \times I_{b}(= V_{p})$ and $Z_{c} \times I_{c}(= V_{c})$ represent the differential and common-mode noise voltage, respectively, placed in the patient's body from some of the scalpel current flowing from the active electrode, through the patient, to the return plate of the ESU. They are related to the location of the ECG electrodes; $V_{p}$ can be reduced by arranging the vector of the differential input ECG electrode perpendicularly to the scalpel current vector. Here, we disregard their contributions since the impedances of the body ($Z_{b}, Z_{c}$) seem to be very small (HUHTA and WEBSTER, 1974) and are not under our control. The last term represents the conversion from the common-mode noise voltage $V_{cm}(= Z_{ec} \times I_{o})$ to the differential noise voltage resulting from the common-mode rejection ratio (CMRR) and the impedance unbalances present in the system.

Because usually:

$$Z_{cm} = Z_{cm1} = Z_{cm2} \approx Z_{e1}, Z_{e2}$$

and

$$Z_{d} = Z_{e1} - Z_{e2}$$

(3)

Eqn. 1 is then reduced to

$$V_{n} = \frac{V_{p}}{K} + Z_{ec}I_{o}\left[ \frac{1}{CMRR} + \frac{Z_{d}}{Z_{cm}} \right]$$

(4)

which shows that $V_{p}/K$ and $I_{o}$ should, therefore, be decreased to minimise the ESU interference.

It should be noticed that the earth impedance of the monitoring device $Z_{g}$ should be increased to decrease $I_{o}$. Decreased $Z_{ec}$ and $Z_{d}$ and increased CMRR and $Z_{cm}$ also reduce the ESU interference.

### 3 Measurement of the common-mode noise voltage

In Fig. 2, the signal between the two electrodes LL and RL is amplified and converted to the optical signal by the battery operated transmitter. This optical signal is transmitted to the receiver via an optical fibre, and the receiver demodulates the original signals. The measuring equipment's bandwidth ranges from 0.05 Hz to 5 kHz, and the common-mode input impedance $Z_{cm}$ is approximately 2 MΩ at 60 Hz. The total impedance $Z_{cm} + Z_{ec}$ between the two electrodes LL and RL is approximately 7 kΩ at 60 Hz. Here, the battery-operated transmitter is isolated completely from the earth and the values of the impedance to the earth $Z_{g}$ can be varied. In this case, $V_{p}/K$ can be ignored because there is no direct conduction pathway from the power line to the transmitter. Still, there may be electric and magnetic induced coupling. This can be minimised by shielding the whole transmitter.

The demodulated signal represents the common-mode noise voltage $V_{ac}$ developed across the RL electrode impedance $Z_{ec}$ due to the HF earth current $I_{o}$ flowing through the RL electrode impedance and $Z_{g}$ to earth. Because the noise voltage developed across the LL electrode impedance is so small (the current flowing through the LL electrode impedance is usually very small compared with $I_{o}$), it can be neglected. The ESU used is a Bovie CSV2, set at 40 (power control) in the cutting mode and is activated to cut the flesh placed on the body.