REDUCING THE IMPEDANCE OF PASSIVE STAINLESS STEEL SURFACE ELECTRODES

Adam W. Wilson[1], BScE, PEng. & Dennis F. Lovely[2], PhD, PEng
1. *Institute of Biomedical Engineering, University of New Brunswick, NB, Canada*
2. *Dept. of Electrical & Computer Engineering, University of New Brunswick, NB, Canada*

INTRODUCTION

Surface electrodes are used in a variety of situations to measure electrophysiological signals from the human body. Most of these measurements, such as the electrocardiogram (ECG), are short-term in nature. However, an area in which surface electrodes are employed chronically is that of powered upper limb prosthetics. In this situation, electrodes are mounted in the socket of the prosthesis to detect the electrical activity of remnant musculature, which in turn is used to control the prosthesis. It is hoped that the prosthesis is worn for most of the day; every day.

It is this application which is addressed in this short paper. A method of reducing the impedance of passive stainless steel electrodes for specific use in myo-electrically controlled prostheses is described. This procedure reduces the susceptibility of the instrumentation to line-borne interference, which can result inadvertent operation of the prosthesis. One of the benefits of the procedure is that it can be applied to existing electrodes and does not require any complicated chemical processes.

BACKGROUND

Warburg first introduced his simple electrode interface model in 1900. This modeled the charge at the interface between the electrode and electrolyte as a capacitor, ***C***, and the half-cell voltage as a battery, $\boldsymbol{E_{hc}}$, with some additional series resistance, $\boldsymbol{R_s}$. Over the years this model has been refined to include some shunting of the interface with a resistive component, $\boldsymbol{R_p}$, as shown in Figure 1. While the schematic of the model remains quite simple, non-linearities arise because of the dependency of the component values on both frequency and current density [1].

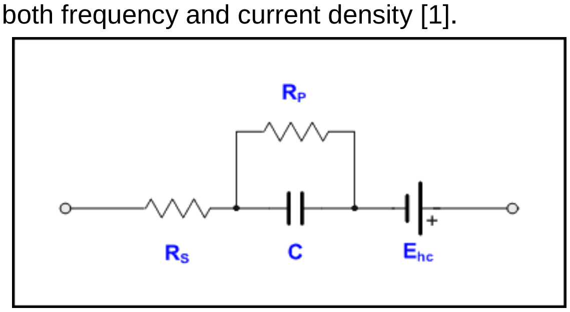

Figure 1: *Modified Warburg electrode model*

When used as surface electrodes, the situation is made more complex with the interaction of the skin layer. It can also be modeled using a similar approach due to the outer layer of the skin being treated as a semi-permeable membrane [2]. The resultant source impedance presented to the 1st stage of an instrumentation system, using electrodes as sensors, is therefore a complex quantity.

EFFECT ON INSTRUMENTATION

The combined impedance of surface electrodes and the underlying tissue vary considerably. Typical magnitudes at 60Hz can range between 5kΩ and 500kΩ, depending on the electrode and hydration level of the skin.

The 1st stage of any electrode signal processing is invariably an instrumentation amplifier with a high common-mode rejection ratio (CMRR). This is necessary to extract the small level biopotential from the ever present common-mode (CM) line-borne interference that is capacitively coupled to the body. However, even if the CM performance of the IA is ideal, differences in source impedance can still give rise to considerable contamination of the detected signal. For a common-mode voltage on the body, $\boldsymbol{E_{CM}}$, and a difference in source impedance, $\boldsymbol{|Z_{e1}-Z_{e2}|}$, the contamination can be expressed as:

$$v_o \approx A_d \cdot \frac{|Z_{e1} - Z_{e2}|}{R_{id}} \cdot E_{CM} \qquad \text{...Eqn. 1}$$

…where $\boldsymbol{A_d}$ is the differential gain and $\boldsymbol{R_{id}}$ is the input resistance of the instrumentation amplifier. To minimize this interference it is important to have $\boldsymbol{R_{id}}$ large and/or $\boldsymbol{|Z_{e1}-Z_{e2}|}$ small.

ELECTRODES FOR MYOELECTRIC CONTROL

In a clinical setting, such as in the recording of an an ECG, silver-silver chloride (Ag-AgCl) electrodes are used due to their stable half-cell potential, low impedance and insensitivity to motion artifact. This last characteristic refers to the generation of small voltages as the electrode is moved relative to the surface of the skin. These electrodes, which are chemically active, are used with a conductive paste which when coupled with good skin preparation provide a low-impedance sensor that minimizes $\boldsymbol{|Z_{e1}-Z_{e2}|}$. This coupled with instrumentation with very high input impedance, results in low levels of line-borne interference.

However, such electrodes are often not suitable for long-term chronic use due to the effects of the paste drying and the possibility of skin irritation. With myoelectric control, the electrodes are generally an inert material, such as gold or stainless steel. They are used dry, without paste or skin preparation, which make the prostheses more acceptable to the user.

Unfortunately, such electrodes, while suitable for chronic applications, do not have the same electrical properties of the Ag-AgCl types. The impedance of such electrodes tends to be much higher than that of similar sized Ag-AgCl types. Similarly, they are very susceptible to motion artifact. To mitigate the susceptibility to motion artifact it has been proposed that the input impedance of the instrumentation be lowered to allow rapid charge redistribution on the surface of the skin [3]. Therefore, the only way to reduce susceptibility to line-borne interference is to closely match the electrode impedances, Z_{e1} & Z_{e2}. From Eqn. 1, it should be noted that the level of CM interference is directly proportional to the difference of these impedances. Consequently, reduction of both of these values should also result in a reduction of the difference.

ELECTRODE SURFACE MODIFICATION

The subject of electrode interface impedance reduction has been tackled in many different ways in the past. Electrochemically chloriding of silver electrodes is one such procedure. However, the optimum chloriding thickness for minimum impedance [4] is quickly removed through cleaning. Aqua regia[1] etching of the electrode surface after heavy chloriding has also been used [5], but both these techniques are not appropriate for electrodes in chronic use.

With the development of implantable micro-electrodes for neural prostheses the situation has been exacerbated. The extremely small cross-sectional area of these electrodes, leads to impedances at 60Hz in the order of 10MΩ or higher. A recent approach to improve this situation has been to use a low-pressure chemical vapour deposition technique to put a rough polysilicon layer on a substrate, prior to evaporating a gold conductive layer [6]. This procedure increased the effective surface area of the gold electrode and lowered the impedance by factor of 45.

We have taken this concept of increasing the effective surface area of micro-electrodes and applied this to the macroscopic world. The simple idea is to sandblast the stainless steel electrodes, commonly used in myoelectric control systems. The hypothesis being that this will pit the smooth surface and so increase the effective surface area. In this way the impedance will be lowered and the susceptibility to line-borne interference reduced. An advantage of this approach is that this surface modification can be applied to existing electrodes and does not have to be part of the manufacturing process.

INITIAL EXPERIMENTATION

The first steps taken to test this hypothesis was to sandblast two pairs of polished stainless steel electrodes. This was done using a BNP 2C-28 blast cabinet fed from a 100 psi air line. The electrodes were held approximately 50mm from the nozzle and one pair sandblasted for 10s, while the other pair was exposed to the sand for 60s. The test electrodes comprised of three pairs of electrodes, namely:

Set **A**: Polished stainless steel - untreated
Set **B**: Lightly sandblasted – 10s
Set **C**: Heavily sandblasted – 60s

The three pairs of electrodes were mounted on a cloth strap to form three bipolar pairs with a centre to centre spacing of 20mm. All three pairs where held on the surface of the forearm and impedance measurements conducted using a previously reported technique [7].

Figure 2: *Three sets of test electrodes*

This technique consists of driving the electrodes under investigation from a sinusoidal source through a known fixed resistor, R_x. By measuring the voltages at each side of the resistor, V_1 and V_2, along with the associated phase angle, φ, the complex impedance of the electrode-skin interface can be determined as:

$$Z_e = \frac{V_2}{2I_{Rx}} = \frac{R_x}{2}\left(\frac{V_2\cos\phi + jV_2\sin\phi}{V_1 - V_2\cos\phi - jV_2\sin\phi}\right) \quad \text{...Eqn. 2}$$

If the tissue impedance between the electrodes is considered small compared to that of the electrode-skin interface, then the components of the modified Warburg model can be extracted from a Bode magnitude plot via minimization of the mean square error between the model and measured data.

[1] A highly corrosive mixture of nitric and hydrochloric acid.

PRELIMINARY RESULTS

Three trials were performed on the same subject on two different days. In all cases the electrode capacitance, ***C***, increased between the polished and lightly sandblasted electrodes. The corresponding $\boldsymbol{R_p}$ magnitude had a significant decrease ($p<0.05$), supporting the hypothesis that sandblasting had increased the effective surface area of the electrodes. However, this trend was not continued for the heavy sandblasting case.

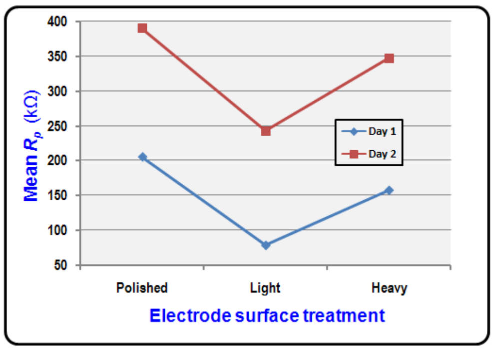

Figure 3: *Variation of* $\boldsymbol{R_p}$ *with surface treatment*

With heavy sandblasting, the average value of $\boldsymbol{R_p}$ increased to a value that was not significantly lower than the untreated electrode. It was also noted that there was substantial variation in values of $\boldsymbol{R_p}$ from day-to-day, as shown in Figure 3.

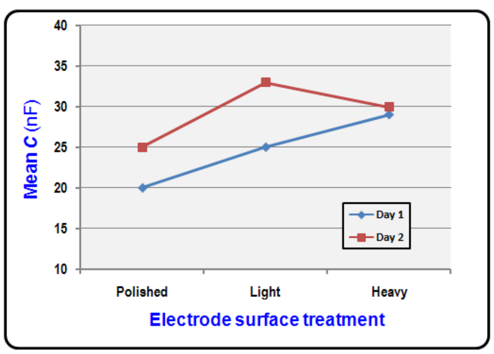

Figure 4: *Variation of **C** with surface treatment*

The value of electrode capacitance, ***C***, generally showed a trend of increase with sandblasting intensity with day-to-day variations not as dramatic as the values of $\boldsymbol{R_p}$.

While the variation in the model element values indicates that the surface treatment by sandblasting is having the desired effect, the important measure for the intended application is the impedance magnitude at 60Hz. The average values of the electrode-skin impedance measured at 60Hz are shown in Table 1 for one representative test set.

This shows a similar trend to that of the electrode interface model elements in which the lightly sandblasted electrode had the best reduction in impedance.

Surface treatment	**|Z| @ 60 Hz**	**Δ|Z|**
Polished (A)	61.1kΩ	
Light sandblast (B)	43.5kΩ	-29%
Heavy sandblast (C)	52.6kΩ	-14%

Table 1: *Average impedance magnitudes at 60Hz*

MICROSCOPY

To try and ascertain the reason behind this somewhat anomalous behaviour, photographs of the surface of the electrodes were obtained using a scanning electron microscope (SEM). It was initially thought that the heavy sandblasting had caused several pits to coalesce into large voids and so circumvent the desired effect of increasing the surface area. However, as Figure 4 shows, this is not the case. The heavily sandblasted electrode does show a rougher surface than the lightly sandblasted one but there is no evidence of large void formation. However, on closer inspection at higher magnification levels it appeared that some of the sand from the heavy sandblasting procedure had become embedded in the stainless steel of the electrode.

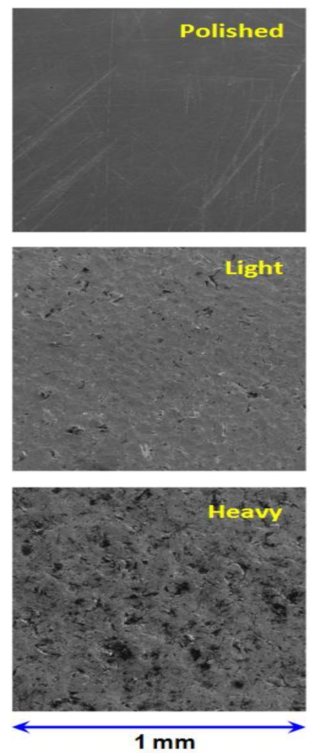

Figure 5: *Electrode surface*

To verify this assumption an EDX (energy dispersive X-ray) analysis was done on the heavily sandblasted electrode. This is a process whereby a X-ray spectrum is obtained for every pixel of a SEM image. Element data can then be collated to give a map of the sample showing its composition.

Figure 6 shows an EDX map for silicon alongside the SEM image of a section of the heavily sandblasted electrode. The large pit shown in the lower left of the image (a) looks empty. However, the EDX map (b) shows that the pit is actually filled with silicon from the sandblasting procedure. As sand is an insulator, this is a possible explanation of the increase in the value of

R_p at high levels of sandblasting, while not affecting the value of C as significantly.

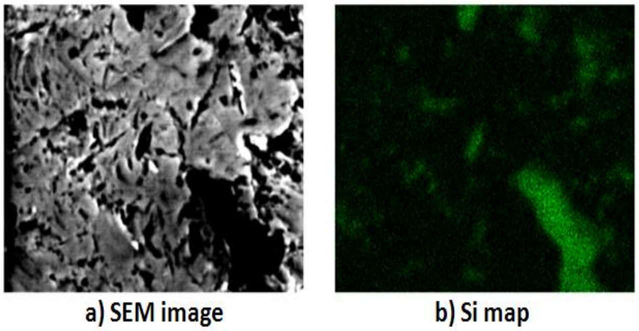

Figure 6: *Silicon embedded in electrode surface*

MATCHING IMPEDANCES

The initial experimentation concentrated on measuring the effect of sandblasting on pairs of stainless steel electrodes. For these experiments it was assumed that both electrodes had identical characteristics. However, as stated earlier, the difference in electrode impedance is a source of CM interference in any electrode instrumentation system.

To measure the difference in impedance between two seemingly identical electrodes a secondary experiment was conducted. The same cloth strap was used to hold the "electrodes-under-test" to the forearm and a common reference electrode, was placed on the back of the hand. The combined impedance comprising of one test electrode, Z_e, body impedance, Z_b, and reference electrode impedance, Z_{ref} was then measured. By doing two measurements, one on each test electrode, while keeping the reference electrode undisturbed, an estimate of the difference in impedance between the two test electrodes can be determined.

RESULTS

Table 2 shows the result of this secondary experiment. The untreated electrodes exhibited a considerable impedance difference. This was greatly reduced by sandblasting the surfaces.

Surface treatment	Δ|Z| @ 60 Hz
Polished (A)	15.4kΩ
Light sandblast (B)	3.4kΩ
Heavy sandblast (C)	3.0kΩ

Table 2: *Matching of impedance magnitudes at 60Hz*.

It is interesting to note that both the lightly and heavily sandblasted electrodes both had good matching characteristics.

CONCLUSIONS

Light sandblasting does lower R_P while increasing the interface capacitance, C significantly. This is consistent with increasing the effective surface area of the electrode. The corresponding magnitude of the impedance at 60Hz is also reduced by almost 30%. Sandblasting also improves the matching between electrodes which should lead to reduced CM interference.

Heavy sandblasting is not as effective a means of lowering electrode impedance values as light sandblasting. This is possible due to the embedding of sand into the surface of the stainless steel.

Day to day variations of electrode interface impedance measurements, even on the same individual, are likely due to the level of skin hydration. Consequently, the value of day-to-day comparisons is limited.

FUTURE WORK

From this preliminary work there appears to be an optimum sandblasting level at which the electrode impedance will reach a minimum. This has yet to be determined. Consequently, further tests using additional levels of sandblasting are planned.

One the optimum level of sandblasting has been determined; the performance of such modified electrodes will be compared to the untreated ones on the basis of line-borne interference.

ACKNOLEDGEMENTS

The authors would like to thank Susan Belfry, director of the Microscopy and Microanalysis Facility at UNB for her assistance with this work. Financial support from NSERC and ACOA is also acknowledged.

REFERENCES

1. Geddes, L. A., Da Costa, C. P., & Wise, G. (1971), "*The impedance of stainless steel electrodes*", Med. & Biol. Eng. , 9, 511-521.
2. Yamamoto, T., & Yamamoto, Y. (1976), "*Electrical properties of the epidermal stratum corneum*", Med. & Biol. Eng. , 14, 151-158.
3. Scott, R.N. & Lovely, D.F (1986), *"Amplifier input impedances for myoelectric control*", Med. & Biol. Eng. & Comp., 24, 527-530.
4. Geddes, LA., Baker, L.E. & Moore, A.G. (1969), "*Optimum electrolytic chloriding of silver electrodes*", Med. & Biol. Eng., 7, 49-56.
5. Grubbs, D.S. & Worley, D.S. (1983), "*New Technique for reducing the imnpedance of silver-silver chloride electrodes"*, Med. & Biol. Eng. & Comput., 21, 232-234.
6. Paik, S.-J., & Cho, D. D. (2002), "*Development of recording microelectrode arrays with low surface impedance for neural chip applications*", J. Korean Phy. Soc. , 41, 1046-1049.
7. Hua, Y., Lovely, D.F. & Doraiswami, R. (2006), *"Factors affecting the stimulus artifact tail in surface recorded somatosensory evoked potentials",* Med. & Biol. Eng. & Comput., 44, 3, 226-241.