Application of the AR analysis approach in order to separate the source and transmission characteristics is the specific contribution of this work and is distinct from other approaches like those by Cohen *et al.* [5]. More extensive experimentation is now needed to further study this model and its clinical application.

References

[1] L. R. Rabiner and R. W. Schafer, *Digital Processing of Speech Signals.* Englewood Cliffs, NJ: Prentice-Hall, 1978.

[2] V. K. Iyer, P. A. Ramamoorthy, Y. Ploysongsang, and H. Fan, "Reduction of heart sounds from lung sounds by adaptive filtering," *IEEE Trans. Biomed. Eng.*, vol. BME-33, pp. 1141–1148, Dec. 1986.

[3] R. L. H. Murphy, Jr., "A simplified introduction to lung sounds," teaching audio tape cassette.

[4] V. K. Iyer, P. A. Ramamoorthy, and Y. Ploysongsang, "Analysis of lung sounds during pulmonary edema in dogs," in *Proc. 8th IEEE/EMBS Conf.*, vol. 2, 1986, pp. 1215–1218.

[5] A. Cohen and D. Landsberg, "Analysis and automatic classification of breath sounds," *IEEE Trans. Biomed. Eng.*, vol. BME-31, pp. 585–590, Sept. 1985.

A Correction Procedure for the Asymmetry of Differential Pressure Transducers in Respiratory Impedance Measurements

RAMÓN FARRÉ, DANIEL NAVAJAS, RENÉ PESLIN, MAR ROTGER, AND CLAUDE DUVIVIER

Abstract—**The usual setup for measuring respiratory input impedance requires a differential pressure transducer attached to a pneumotachograph. As, up to now, no data correction procedure has been devised to account for transducer asymmetry, a highly symmetrical transducer is required to obtain reliable impedance data. In this communication, a general model for the measuring system is presented. Its main feature is that differential pressure transducers are modeled as two input–one output systems. From the theoretical model, we defined a dynamic calibration and data correction procedure. This was tested using highly asymmetrical transducers (common-mode rejection ratio between 45 and 27 dB) to measure the impedance of two respiratory analogs. The latter were linear resistance (R), inertance (I), compliance (C) series models simulating a normal subject ($R = 3.47$ hPa · s · l^{-1}, $I = 1.45$ Pa · s^2 · l^{-1}, $C = 18.6$ ml · hPa^{-1}) and an obstructive patient ($R = 11.15$ hPa · s · l^{-1}, $I = 1.28$ Pa · s^2 · l^{-1}, $C = 18.5$ ml · hPa^{-1}). Results obtained applying the devised procedure (errors in R, I, and C always less than 4 percent) show that respiratory input impedance can be adequately measured if data are corrected for transducer asymmetry.**

Introduction

Respiratory mechanical impedance (Z_{RS}) is commonly obtained by connecting the subject's airway to a pressure generator, and by relating airway pressure to the pressure drop (ΔP) across a reference impedance (pneumotachograph Z_{PN}) placed in series with the subject. To avoid interfering with the subject's breathing, Z_{PN} is usually small compared to Z_{RS}. It follows that ΔP, as measured with a differential pressure transducer, is but a small fraction of the common pressure applied to both inputs of this transducer. An accurate estimate of Z_{RS} therefore requires a high dynamic symmetry of the transducer, as defined by its common-mode rejection ratio (CMRR): CMRR is defined as CMRR $= 20 \cdot \log (P/\Delta P')$ where $\Delta P'$ is the spurious differential pressure recorded by the transducer when the same pressure P is simultaneously applied to both inputs of the transducer. It has recently been shown that, with common pneumotachographs, substantial errors are made in respiratory parameters when CMRR is less than 60 dB at 30 Hz, particularly when Z_{RS} is high [5]. So far, transducer asymmetry, which increases with frequency, has been viewed as an insuperable circumstance [2], [5], and no correction scheme has been devised to eliminate the corresponding error. Indeed, transducers have always been modeled as one input–one output systems, with a single frequency response [2], [7], while dynamic asymmetry can only be accounted for by a two input–one output model. As a consequence, accurate impedance data can only be obtained on a limited frequency range using the most symmetrical transducers presently available.

In this paper, we present a general theoretical model of the most usual impedance measuring system taking into account the asymmetry of differential pressure transducers. From this model, we propose a simple dynamic calibration procedure which allows correction of impedance data for transducer asymmetry.

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Theory

Fig. 1(a) shows a diagram of the setup generally used to measure respiratory input impedance. All magnitudes are in the frequency domain and their dependence on frequency is omitted. The excitation pressure generated by the loudspeaker is recorded at the mouth (P_E) with a differential pressure transducer (PT1). The excitation flow ($\dot{V}_E$) entering the respiratory system is measured as the pressure drop ($\Delta P = P_1 - P_2$) across a pneumotachograph (PN) by means of another differential pressure transducer (PT2). Thus, from the recorded signals S_P and S_V, a measured impedance value Z_M

$$Z_M = S_P / S_V \tag{1}$$

is calculated for the true respiratory impedance Z_{RS} defined as

$$Z_{RS} = P_E / \dot{V}_E. \tag{2}$$

The mouthpiece is, in fact, included in the respiratory system. Its small influence is eliminated by correcting Z_{RS} for the known impedance of the mouthpiece.

The main feature of the following theoretical approach is to consider a differential transducer as a linear system with two inputs and one output [Fig. 1(b)]. Thus, a transducer is characterized by two associated transfer functions [1] corresponding to the positive and negative pressure ports: H_P^+ and H_P^- for PT1, H_V^+ and H_V^- for PT2. Each of these transfer functions accounts for the relationship between the signal provided by the transducer and the corresponding pressure applied to one of its ports while the other is maintained at the atmospheric ground level. Therefore, from the block diagram in Fig. 1(b), the recorded signals S_V and S_P are

$$S_V = P_1 H_V^+ + P_2 H_V^-$$
$$S_P = P_E H_P^+. \tag{3}$$

S_P does not depend on H_P^- since the negative port of PT1 is open to the atmosphere. This does not mean that the response of PT1 is unaffected by the dimensions of the tubing attached to the negative port. In fact, as occurs with a compliant-membrane differential pressure transducer with connecting tubes [5], the transfer function

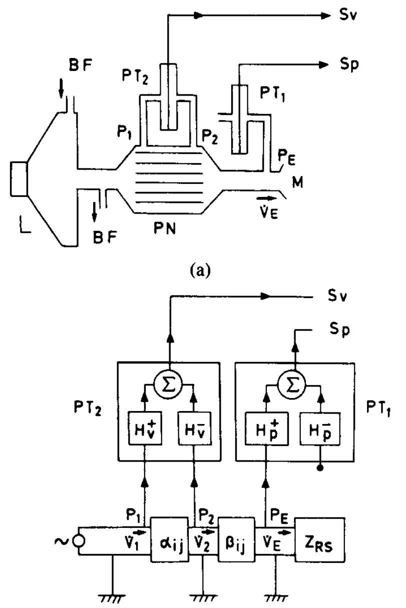

Fig. 1. (a) Diagram of a respiratory impedance measurement setup: loudspeaker (L), bias flow (BF), pneumotachograph (PN), mouthpiece (M), pressure transducers (PT1, PT2) recorded signals (S_V, S_P). (b) General model including block diagrams of transducers and recorded signals.

corresponding to the positive port (i.e., H_P^+) depends on the dimensions of the tubing connected to the positive port as well as on the tubing connected to the negative port.

Moreover, the pneumotachograph and the tube connecting the outlet of the pneumotachograph to the point where excitation pressure is measured as well as the input impedance of pressure transducers [3] are considered as general two-port systems [Fig. 1(b)] characterized by their impedance coefficients α_{ij} and β_{ij} ($i, j = 1, 2$) [6]. Thus, the involved pressures (P_1, P_2, P_E) and flows ($\dot{V}_1$, $\dot{V}_2$, $\dot{V}_E$) are related by the following equations:

$$\begin{aligned} P_1 &= \alpha_{11}\dot{V}_1 - \alpha_{12}\dot{V}_2 \\ P_2 &= \alpha_{12}\dot{V}_1 - \alpha_{22}\dot{V}_2 \\ P_2 &= \beta_{11}\dot{V}_2 - \beta_{12}\dot{V}_E \\ P_E &= \beta_{12}\dot{V}_2 - \beta_{22}\dot{V}_E. \end{aligned} \tag{4}$$

After algebraic combination of (1)-(4), the following equation relating Z_{RS} and Z_M is obtained:

$$Z_{RS} = A(1/Z_M - B)^{-1} \tag{5}$$

where

$$A = \frac{H_V^+}{H_P^+}\left[\frac{\alpha_{11}\alpha_{22} - \alpha_{12}^2}{\alpha_{12}}\frac{\beta_{22}}{\beta_{12}} + \frac{\beta_{11}\beta_{22} - \beta_{12}^2}{\beta_{12}}\frac{\alpha_{11}}{\alpha_{12}}\right] + \frac{H_V^-}{H_P^+}\left[\frac{\beta_{11}\beta_{22} - \beta_{12}^2}{\beta_{12}}\right] \tag{6}$$

and

$$B = \frac{H_V^+}{H_P^+}\left[\frac{1}{\beta_{12}}\frac{\alpha_{11}\alpha_{22} - \alpha_{12}^2}{\alpha_{12}} + \frac{\alpha_{11}}{\alpha_{12}}\frac{\beta_{11}}{\beta_{12}}\right] + \frac{H_V^-}{H_P^+}\frac{\beta_{11}}{\beta_{12}}. \tag{7}$$

Apart from their implicit dependence on frequency, A and B only depend on the mechanical characteristics of the measuring device and on the transfer functions of transducers. Therefore, (5) allows us to design a simple dynamic calibration procedure to obtain Z_{RS} from Z_M. First, B can be calculated from the obtained impedance (Z_M^∞) when the measuring system is occluded ($Z_{RS} = \infty$):

$$B = 1/Z_M^\infty. \tag{8}$$

Second, A can be calculated from the obtained impedance (Z_M^{REF}) when a known reference impedance (Z_{REF}) is attached to the measuring device. From (5) and (8), we obtain that

$$A = Z_{\mathrm{REF}}(1/Z_M^{\mathrm{REF}} - 1/Z_M^\infty). \tag{9}$$

Thus, combining (5), (8), and (9), we finally get

$$Z_{RS} = Z_{\mathrm{REF}}\frac{1/Z_M^{\mathrm{REF}} - 1/Z_M^\infty}{1/Z_M - 1/Z_M^\infty}. \tag{10}$$

The interest of (10) is that by performing two simple calibration measurements, it is possible to calculate Z_{RS} from the measured value Z_M. Only a known reference impedance is required.

The above general equations can be simplified in the particular case where: 1) gas compression inside the pneumotachograph is negligible, i.e., the pneumotachograph can be represented by its series impedance Z_{PN} ($Z_{PN} = (\alpha_{11}\alpha_{22} - \alpha_{12}^2)/\alpha_{12}$); 2) the excitation pressure is measured at a point sufficiently close to the outlet of the pneumotachograph so that we can neglect the system characterized by β_{ij}; and 3) an analog or digital filter is used to compensate for any difference in the frequency responses of transducers and of the pneumotachograph, i.e., $(1/Z_{PN})(H_P^+/H_V^+) = 1$. Under these special conditions, it follows from the previous equations that

$$\frac{Z_M}{Z_{RS}} = \left[1 + \frac{Z_{RS}}{Z_{PN}}\frac{H_V^+ + H_V^-}{H_V^+}\right]^{-1} \tag{11}$$

and (10) is reduced to

$$Z_{RS} = (1/Z_M - 1/Z_M^\infty)^{-1}. \tag{12}$$

Equation (11) shows that even in this case, an error in the estimation of impedance would remain ($Z_M \neq Z_{RS}$) due to the asymmetry of the differential pressure transducer ($H_V^+ \neq -H_V^-$). As illustrated in Fig. 2, this error depends on the degree of asymmetry characterized by means of the CMRR and on the ratio Z_{RS}/Z_{PN}. It appears that in the absence of any other source of error, the asymmetry of the transducer can produce errors up to 50 percent for a ratio Z_{RS}/Z_{PN} around 30 and a CMRR = 30 dB. These conditions can be found in practice when measuring high impedances as in patients or in children [8] with common pressure transducers. Equation (12) shows that under the above-mentioned conditions, the calibration procedure becomes even easier since no reference impedance is required.

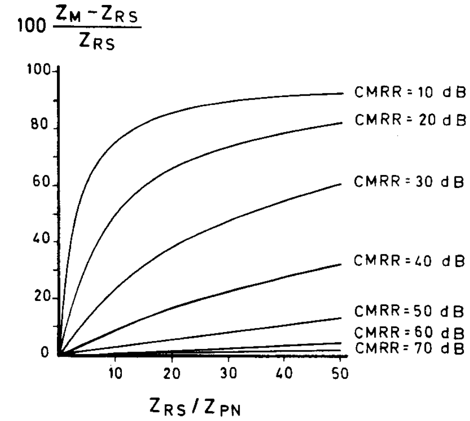

Fig. 2. Error due to transducer asymmetry in impedance measurements for different values of Z_{RS}/Z_{PN} and CMRR. All magnitudes in (11) are assumed to be real and CMRR is approximated by $20 \cdot \log(H_V^+/(H_V^+ + H_V^-))$.

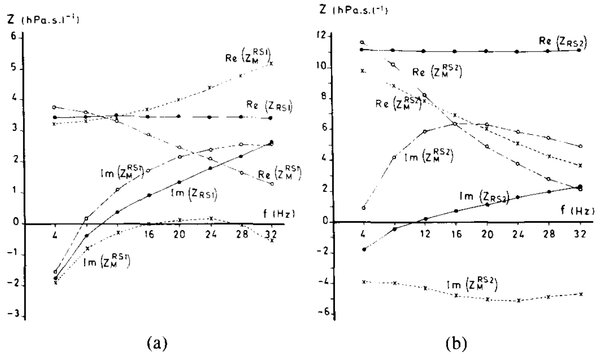

Fig. 3. (a) Real part (Re) and imaginary part (Im) of the measured impedance of $RS1$ analog (Z_M^{RS1}). Impedances plotted with (×) and with open circles correspond to data obtained with both reversed positions of transducer $PT2$. The true impedance of the analog (Z_{RS1}) is drawn with closed circles. (b) Same for analog $RS2$.

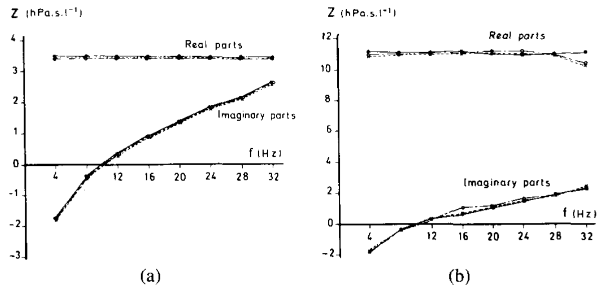

Fig. 4. (a) Real part (Re) and imaginary part (Im) of the recovered impedance of the respiratory system analogs after correction of data in Fig. 3 using (10). Impedances plotted with (×) and with open circles correspond to data obtained with both reversed positions of transducer $PT2$. The true impedance of the analog (Z_{RS1}) is drawn with closed circles. (b) Same for analog $RS2$.

Measurements

The correction procedures suggested by (10) and (12) were tested under extreme conditions: measuring a high impedance using transducers with a very low CMRR. To do this, we used two linear series resistance (R), inertance (I), compliance (C) mechanical analogs of the respiratory system. They were constructed as described by Peslin *et al.* [5]. The first analog ($RS1$) represented a normal subject ($R = 3.47$ hPa · s · l^{-1}, $I = 1.45$ Pa · s^2 · l^{-1}, $C = 18.6$ ml · hPa^{-1}), and the second one ($RS2$) roughly simulated the impedance of an obstructive patient ($R = 11.15$ hPa · s · l^{-1}, $I = 1.28$ Pa · s^2 · l^{-1}, $C = 18.5$ ml · hPa^{-1}). Similarly, a reference impedance (REF: $R = 3.35$ hPa · s · l^{-1}, $I = 0.17$ Pa · s^2 · l^{-1}) to be used in the calibration was constructed. The impedance of these analogs (Z_{RS1}, Z_{RS2} and Z_{REF}) was accurately measured with sinusoidal excitation (± 1 hPa) by: 1) using highly symmetrical pressure transducers (Validyne, MP-45) with a CMRR greater than 70 dB up to 32 Hz, 2) compensating the frequency response of the pneumotachograph (Fleish II) with a first-order transfer function ($\tau = 2.2$ ms [4]), 3) carefully matching H_P^+ and H_V^+, and 4) correcting for the shunt gas compression inside the pneumotachograph and the connecting tube. Impedances were calculated using the FFT algorithm [5]. To verify that no error was induced by transducer asymmetry, the measurements were repeated after interchanging the pressure ports in PT2: the greatest difference found was 2 percent for the real part of Z_{RS2} at 32 Hz.

Once Z_{RS1}, Z_{RS2}, and Z_{REF} were known, the two Validyne transducers were replaced by two identical Statham PM-283 pressure transducers. The latter are asymmetrical due to the different volumes of the chambers on both sides of the membrane. Frequency responses H_P^+ and H_V^+ of both transducers were matched. Their CMRR decreased from 45 dB at 4 Hz to 27 dB at 32 Hz. Taking into account the pneumotachograph impedance Z_{PN} ($R = 0.30$ hPa · s · l^{-1}, $I/R = 2.2$ ms), the quotient Z_{RS}/Z_{PN} was around 12 and 37 for $RS1$ and $RS2$, respectively, over most of the frequency range. Thus, a large error in impedance estimation could be expected (Fig. 2). Finally, Z_M^∞, Z_M^{REF}, Z_M^{RS1}, and Z_M^{RS2} were measured. These measurements were also carried out with the pneumotachograph's transducer inputs connected in the reversed sense.

Fig. 3(a) depicts the true impedance (Z_{RS1}) of analog $RS1$ and the corresponding measured values Z_M^{RS1} for both reversed senses of transducer PT2. Similarly, the same data concerning analog $RS2$ are shown in Fig. 3(b). As expected, greater errors were found, especially as frequency increased (as CMRR decreased). The errors were not negligible even at 4 Hz for the normal respiratory system analog $RS1$ ($Z_{RS}/Z_{PN} = 12$ and CMRR = 45 dB). Due to the different role that H_V^+ and H_V^- play in (6) and (7), different results were obtained when reversing the position of transducers. As shown in Fig. 4(a) and (b), when the devised procedures were applied to correct Z_M^{RS1} and Z_M^{RS2}, we recovered the true impedance of analogs with errors in R, I, and C less than 4 percent when using both the simplified equation (12) and the general equation (10).

Discussion

The devised data correction procedure can be widely used since the general model in Fig. 1(b) is based on a linear analysis. This can be assumed for the involved systems (common pressure transducers, pneumotachograph, and connections) taking into account the range of pressure and flow signals during respiratory impedance measurements. Moreover, no other hypothesis has been assumed for the transducer response or for the dynamic behavior of the pneumotachograph and connections. This fact is of special interest at high frequencies where air compression inside these parts of the measuring system could become important.

The dynamic calibration procedure to correct measured impedance data is very easy to carry out in practice since only two normal impedance measurements are needed: with the measuring system occluded and when a known reference impedance is connected to the system. The last measurement can be avoided provided the previous simplifying hypotheses are met. An interesting feature of the procedure is that the calibration measurements are carried out with all the elements of the measuring system in their place, i.e., without disassembling transducers, thus avoiding errors associated with changes in their connections. In this way, it is not necessary to know separately each of all the magnitudes appearing on the right of (6) and (7), but only their two global combinations: A and B parameters. Therefore, A and B globalize all the physical information necessary for the correction. Apart from its practical simplicity, calculation of A and B from (8) and (9) instead of directly applying (6) and (7) should reduce measurement errors since in the first case, only two values must be measured (Z_M^∞ and Z_M^{REF}), while in the second one, more than two magnitudes must be measured or matched.

Measurements in this work were carried out using a common pneumotachograph. To determine the true impedance of analogs, the most symmetrical transducers currently available [5] were used and frequencies were restricted below 32 Hz to guarantee a CMRR greater than 70 dB and thus negligible errors (Fig. 2). The pressure transducers used to test and illustrate our method presented a substantial, but in no way exceptional degree of dynamic asymmetry (27 dB at 32 Hz). In the measurements on analog $RS2$ at the highest frequencies, the spurious pressure due to transducer asymmetry was even bigger than the differential pressure to be measured: $10^{-\text{CMRR}/20} = 3.2$ percent and $Z_{PN}/Z_{RS} = 2.7$ percent of the common pressure supported by both inputs, respectively. This degree of asymmetry is clearly inadequate for measuring respiratory impedance above a few hertz, particularly in patients [Fig. 3(b)], unless the data are corrected by our procedures. We obtained good results even when Z_{RS}/Z_{PN} was around 37 and CMRR was less than 30 dB, that is, much lower than the value of 70 dB which has been shown to be acceptable in the range of 4–30 Hz with common pneumotachographs [5].

As expected, we obtained similar results when applying (10) and (12) since the hypotheses assumed for the simplification were met during our measurements. First, the frequency responses of transducers, pneumotachograph, and filters were compensated. Second, the volume of air inside the pneumotachograph was only 11 cm^3, giving gas compression negligible up to 32 Hz. Third, the tube connecting the outlet of the pneumotachograph with the point where excitation pressure was measured had a length of 4 cm and an internal diameter of 2.8 cm allowing us to neglect β_{ij}. Nevertheless, application of the simplified correction, (12), could not be used under different circumstances, for instance, at high frequencies where shunt impedances play a larger role.

In conclusion, we established a general model for the most common setup used to measure respiratory input impedance. From this theoretical basis, we also devised a simple calibration and data correction procedure. The latter was shown to adequately correct very large errors due to transducer asymmetry. With this approach, even highly asymmetrical differential pressure transducers can be used to measure respiratory input impedance. This is already of interest in the usual frequency range. Moreover, it permits measurements at higher frequencies where a high enough common-mode rejection ratio of the differential pressure measuring system is difficult to obtain.

Acknowledgment

The authors are grateful to M. C. Rohrer for the illustrations.

References

[1] J. S. Bendat and A. G. Piersol, *Random Data: Analysis and Measurement Procedures.* New York: Wiley-Interscience, 1971.

[2] E. Delavault, G. Saumon, and R. Georges, "Identification of transducer defect in respiratory impedance measurements by forced random noise. Correction of experimental data," *Resp. Physiol.*, vol. 40, pp. 107–117, 1980.

[3] R. Farré, R. Peslin, D. Navajas, C. Gallina, and B. Suki, "Analysis of the dynamic characteristics of pressure transducers for studying respiratory mechanics at high frequencies," *Med. Biol. Eng. Comput.*, vol. 27, 1989.

[4] R. Peslin, J. Morinet-Lambert, and C. Duvivier, "Etude de la réponse en fréquence de pneumotachographes," *Bull. Physio-Pathol. Resp.*, vol. 8, pp. 1363–1376, 1972.

[5] R. Peslin, P. Jardin, C. Duvivier, and P. Begin, "In-phase rejection requirements for measuring respiratory input impedance," *J. Appl. Physiol.*, vol. 56, no. 3, pp. 804–809, 1984.

[6] R. Peslin and J. J. Fredberg, "Oscillation mechanics of the respiratory system," in *Handbook of Physiology, Vol. 3, Part I.* Bethesda, MD: Amer. Physiol. Soc., 1986, sect. 3.

[7] R. L. Pimmel, R. A. Sunderland, D. J. Robinson, H. B. Williams, R. L. Hamlin, and P. A. Bromberg, "Instrumentation for measuring respiratory impedance by forced oscillation," *IEEE Trans. Biomed. Eng.*, vol. BME-24, pp. 89–93, 1977.

[8] S. P. Williams, J. M. Fullton, M. J. Tsai, R. L. Pimmel, and A. M. Collier, "Respiratory impedance and derived parameters in young children by forced random noise," *J. Appl. Physiol.*, vol. 47, pp. 169–174, 1979.

Comments on "A Spiral Nerve Cuff Electrode for Peripheral Nerve Stimulation"

GABOR B. RACZ AND JAMES E. HEAVNER

Introduction

We are of the opinion that cuff electrodes for peripheral nerve stimulation have little future. Our poor prognosis is based upon the history of this type of electrode, and it has not been changed by the optimistic report by Naples *et al.*[1] On the contrary, we feel that electrodes similar to or exactly like those used for epidural spinal cord stimulation can be used for peripheral nerve stimulation. Problems of nerve irritation are eliminated by putting a thin flap of fascia between the electrode and the nerve [1], [2]. In this regard, this is quite similar to stimulating the spinal cord through the dura. Availability of four contact points through which to stimulate and the availability of programming a stimulation pattern, and the ability to change this stimulation pattern, offer distinct advantages. Obviously, such an approach does not allow one to stimulate individual axons but neither will a cuff technique. We have clinical experience with the type of electrode we describe [1], [2]. From this experience we know that we can effectively stimulate nerves and provide pain relief in patients suffering from reflex sympathetic dystrophy/causalgia.

We have recently become aware that one way of providing an animal model of reflex sympathetic dystrophy is to place a loose band around the rat sciatic nerve with suture material [3]. This should be kept in mind when placing stimulating electrodes around a nerve.

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[1] G. G. Naples, J. T. Mortimer, A. Scheiner, and J. D. Sweeney, *IEEE Trans. Biomed. Eng.*, vol. 35, pp. 905–916, 1988.

References

[1] G. B. Racz, T. Browne, and R. Lewis, "Peripheral nerve stimulator for treatment of nerve injury (causalgia) caused by electrical burns. Report of two cases," *Texas Med.*, vol. 84, pp. 45–50, 1988.

[2] G. B. Racz, R. Lewis, J. E. Heavner, and J. Scott, "Peripheral nerve stimulator implant for treatment of causalgia," in *Reflex Sympathetic Dystrophy and Related Syndromes*, M. Stanton-Hicks, Ed. Kluwer Academic, 1988.

[3] G. J. Bennett and Y.-K. Xie, "A peripheral mononeuropathy in rat that produces disorders of pain sensation like those seen in man," *Pain*, vol. 33, pp. 87–107, 1988.

Authors' Reply

G. G. NAPLES, J. T. MORTIMER, A. SCHEINER, AND J. D. SWEENEY

We disagree with the opinions expressed by Dr. Racz and Dr. Heavner regarding our recent paper[1] and we believe that their comments serve little constructive purpose. Racz and Heavner do not criticize any specific aspect of our work; rather, they appear to reference our paper as a means of establishing a forum through which to discuss their own work with a different type of electrode. They express the opinion that "cuff electrodes for peripheral nerve stimulation have little future." They support their opinion by citing the "history" of cuff electrodes. Presumably, this is a reference to the problems of tissue damage and nerve irritation that have sometimes been associated with these electrodes—problems which we summarized clearly in the introduction to our paper. Indeed, Dr. Racz and Dr. Heavner seem to have missed the whole point—that we devised the self-sizing spiral cuff design with the intent of

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