**Publications of Dr. Martin Rothenberg:**

# Airflow-Based Analysis of Vocal Function

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One of the few noninvasive methods available for obtaining a clinically useful estimate or description of the vibratory pattern of the vocal folds is the inverse filtering of either the airflow or pressure waveform at the mouth (Rothenberg, 1973, 1977), that is, the processing of the waveform at the mouth with a filtering system that has a transform approximately the inverse of that of the vocal tract between glottis and lips. For clinical purposes, inverse filtering of the airflow at the mouth rather than the pressure is preferable, since only the airflow method results in a known zero level and an easily calibrated airflow scale for the resulting glottal flow waveform. The circumferentially-vented wire-screen pneumotachograph mask has been shown to yield an oral volume velocity waveform adequate for inverse-filtering up to about 1 kHz to 2 kHz, depending on the mask configuration.

Though such a mask, combined with a manually-adjusted inverse filter, is now being used by many voice research laboratories and a small number of research-oriented clinical facilities, the necessity of properly adjusting the inverse filter parameters for each subject - to match the frequency and damping of the lowest one or two formants (vocal tract resonances) - makes this system impractical for general clinical use. To overcome this problem, a number of laboratories are attempting to develop computer-based, automated inverse filtering algorithms (for example, Javkin, et al., 1987 and Gauffin, et al., 1986). Though of possible value in the long term, presently proposed automated schemes can produce large errors if the program errs. This is likely to occur for grossly abnormal voices, such as highly breathy voices or in the presence of significant nasality. Both of these conditions are counter to the assumptions upon which automated inverse-filtering schemes are normally predicated. Sophisticated schemes for automated inverse-filtering which are robust under a wide variety of voice conditions are yet to be developed.

This paper proposes a system for the airflow-based analysis of vocal function employing a processing scheme for airflow signals that appears to bypass the pitfalls inherent in standard inverse filtering and provide an easily used and robust method for obtaining from the oral airflow waveform those parameters of the glottal waveform having the most significance in clinical applications. The method uses the output of a wide-band circumferentially-vented wire-screen pneumotachograph mask during a spoken vowel having a high first formant, such as /æ/ or /a/ in English, to derive a set of parameters adequate for reconstructing a simplified or stylized version of the glottal waveform. These parameters are:

1. $T_o$, the fundamental period of each cycle of the quasiperiodic flow waveform.
2. $P$, the peak airflow attained during each period $T_o$.

3. *L*, the minimum (lowest) airflow during each period. sometimes referred to as the waveform offset from zero flow.
4. *M*, the mean or average airflow during each period.
5. $Q_o$, often referred to as the open quotient, which is the fraction of each period $T_o$ during which the vocal folds are essentially not in contact.

Each of these variables relates to physiologically significant variables of clinical interest: (1) The importance of $T_o$ as the primary determinant of voice pitch is unquestioned. (2) For a given subglottal pressure. the minimum now *L* indicates the degree to which the vocal folds do not atlain a complele closure during the vibratory cycle. (3) The peak flow *P*, or peak-to-peak flow *P-L*, would correlate well with the peak variation in glottal area and, therefore, with vocal fold mobility and oscillalory efficiency, assuming a given level of subgloltal pressure and ab-adductory force. (4) The mean flow *M* determincs the rate of deflation of the lungs. (5) The open quotient $Q_o$ tends to reflect the degree of abduction or adduction of the vocal folds (as does *P*).

The proposed method is based on certain very general assumptions related to the nature of glottal waveforms, namely, that (1) the primary excitation of the vocal tract resonances for each glottal cycle occurs during thc glottal closing phase, after the occurrence of the peak glottal flow. (2) the vocal tract resonances are more highly damped during the open phase of the glottal cycle, and (3) any strong wavefoml discontinuity in slope - most significantly the abrupt flattening of the waveform caused by the closing of the vocal folds over some portion of their length - will tend to occur near the smaller values of instantaneous airflow rather than the higher values. These assumplions are well supported in the literature and result from the basic physics of vocal fold vibration and vocal tract acoustics. Finally, we assume that for the clinical evaluation of vocal fold vibratory behaviour it is sufficient to record such behaviour during an open vowel, such as /æ/ or /a/.

Under these assumptions, reasonable estimates of the peak and minimum values of the glottal volume velocity waveform can be obtained by measuring the peak and minimum values of low-pass filtered versions of the flow waveform at the mouth. From the first two assumptions it can be inferred that there is little formant energy added to the glottal flow by the vocal tract at the instant of pcak glottal now; thc forrn:lnts would be slimulated just after the pcak now for thc previous glottal cycle. and the resulting energy wouJd have largcly decayed by the time that thc peak now occurs. since the peak flow occurs near, usually just after the instant of maximum glottal area. (See for example Rothenberg, 1973, Figure 16, or Rothenberg, 1977, Figure 8). Thus, a small amount of smoothing or low-pass filtering of the oral waveform, to further reduce formant energy during the glottal open phase, should be sufficient to yield a waveform with a peak value close to that of the glottal waveform. As we have previously shown, a low-pass filter with good phase response and little or no ovcrshoot in its transient response, such as a Bessel-derived filter, can be used for this purpose, if the cutoff frequency of the filter is chosen to be above $F_o$ but significantly below the frequency of the first formant $F_1$ (Rothenberg, 1977, Figure 8).

The minimum value of the glottal waveform is especially well retained by such filtering, since, during the period of relatively constant glottal flow level during a closed phase, there is time for the low-pass filter output to approach this level. For wavefonns with little or no closed phase. The low-pass filtering, as long as it is significantly above $F_o$, will still yield a reasonable minimum value, since the Fourier component at $F_o$ will tend to dominate in both the oral and glottal waveforms.

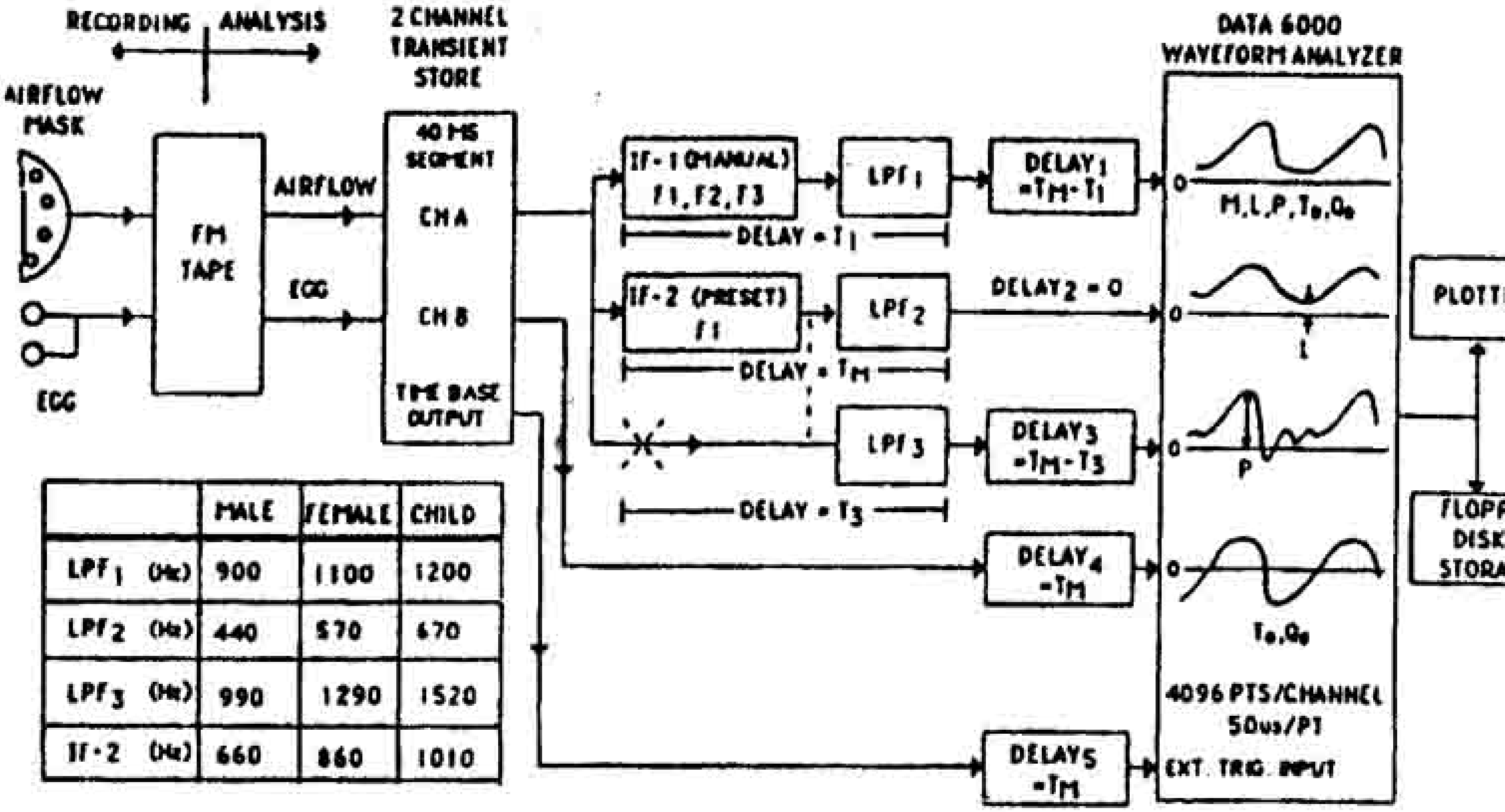
We describe below two implementations of this procedure, as well as initial test results for speakers having a variety of voice qualities. In the implementations to be described. an approximate $F_1$ inverse filter stage was added to the low-pass filtering to increase accuracy with very strong voices, that is, with voices having a relatively high amount of energy at the formant frequencies.

## Method - first experiment

In our first experiment with the newly proposed method, we implemented an automatic parameter measurement system of the type outlined and compared the resulting parameter values with the values obtained by means of a standard inverse-filtering procedure in which the filter parameters are manually adjusted by a trained operator while observing the filtered waveform during a repetitive playback of the voice sample. The system was tested with 29 subjects having a variety of voice qualities.

The test system was implemented on a Data Precision DATA 6000 microprocessor-based waveform analyzer, with some of the signal filtering performed in analog form, before A-D conversion. The system is shown in Figure 1. The output of an airflow mask having a double layer of 500 mesh wire screen and a flow resistance of about 0.5 cm $H_20$/liter per second (Glottal Enterprises model MA-2) and a Laryngograph electroglottograph were recorded on FM tape. The electroglottograph signal was included to allow independent measurements of $T_o$ and $Q_o$, though it was realized that measurements of $Q_o$ derived from airflow and EGG signal could be quite different. The EGG signal was also used occasionally as an indication of the glottal closed period in setting the manual inverse filter parameters (Rothenberg, 1979).

During analysis, a 40 msec segment of each vowel to be tested was first captured on a two-channel, wide-bandwidth transient storage unit. This segment was then recorded in the DATA 6000 signal analyzer in four forms:
(1) On Channel 1, a manually inverse-filtered glottal waveform was recorded, using a standard analog filter (Glottal Enterprises model MSIF). Though four formants could be removed by this filter, only three zero pairs (antiresonances or antiformants) had any noticeable effect on the waveform for the voices tested.
(2) On Channel 2. an airflow signal was recorded that was passed through a single formant approximate inverse filter set for the average first formant for the vowel /æ/ for adult males, adult females or children, depending on the subject, as taken from the classical study by Peterson and Barney (1952). The anti formant (complex zero) damping factor was set to zero, though it was later determined that a setting of about 0.5 in damping factor would have led to slightly more accurate values of $L$ in some cases. An 8-pole Bessel low-pass filler with -3dB cutoff frequency set at 2/3 times the average formant frequency for that subject category (Male, Female or Child) was also used to funher attenuate the formant energy, as required by the proposed system design for estimating the minimum glottal airflow parameter $L$. The Channel 2 signal was also used by the DATA 6000 for estimating the waveform period $T_o$ and the mean airflow M.
(3) On Channel 3. the airflow waveform was only slightly low-pass filtered, using an 8-pole Bessel filter set to -3dB at the relatively high value of 1.5 times the average $F_1$ for the subject-age calcgory. According to the system design, the maximum of this signal during $T_o$ would be used Cor estimating the peak glottal airflow $P$.
(4) The EGG waveform was recorded on Channel 4.




| | MALE | FEMALE | CHILD |
|---|---|---|---|
| LPF1 (Hz) | 900 | 1100 | 1200 |
| LPF2 (Hz) | 440 | 570 | 670 |
| LPF3 (Hz) | 990 | 1290 | 1520 |
| IF-2 (Hz) | 660 | 860 | 1010 |

Fig. 1. Analysis system used. The dashed lines show changes for the second expe ment. Settings for LPF3 in the second experiment were changed to 660 Hz (Male), 8 Hz (Female), and 1010 Hz (Child). The time delay units were adjusted to put all wa forms in correct time alignment.

A program on the DATA 6000 automatically derived $T_o$, $M$, $L$, $P$ and $Q_o$. To was measured at a criterion level approximately half way between thc maximum and minimum values of the captured sample in channel 2, and $M$ was computed as the mean of all data points in the channel 2 waveform during the period $T_o$, $L$ and $P$ were measured according to the rules indicated in Figure 1.

The open quotient $Q_o$ was estimated from the airflow parameters $P$, $L$ and $M$ by assuming a model for the glottal waveform of a sinusoid truncated at its lower extreme. According to this model, $Q_o$ is uniquely relaled to $P$, $L$ and $M$ by the equation:

$$\sin(\pi Q_o) - \pi Q_o \cos(\pi Q_o) / 1 - \cos(\pi Q_o) = \pi M / P - L$$

We found this equation to yield a reasonable first approximation for $Q_o$, given accurate estimates of $P$, $L$ and $M$.

The system in Figure 1 was tested using 29 subjects as follows:
6 normal adult males
6 dysfunctional adult males

6 normal adult females
3 dysfunctional adult females
7 normal children (5 female and 2 male, 7 to 13 years old)
1 dysfunctional child (male, 11 years old).

The dysfunctional adults included cases of laryngitis, diplophonia secondary to laryngitis, Parkinson's disease, post-surgery-trauma-induced left vocal fold paralysis, trauma-induced breathiness, and simulated hyperfunctional-adducted phonation. The child's vocal dysfunction was caused by a vocal fold nodule. Each subject was asked to vocalize a short held /æ/ at a normal conversational level, and at levels roughly 6dB above and below this level, as monitored by the subject on a digital (LED) level display. The subject's most comfortable pitch was used at each level. Twenty-eight subjects produced 3 loudness levels and 1 subject produced 4 loudness levels, resulting in a total of 88 data points. The manual inverse-filtering was performed by the second author or a graduate research assistant, with each previously trained in this task by the first author.

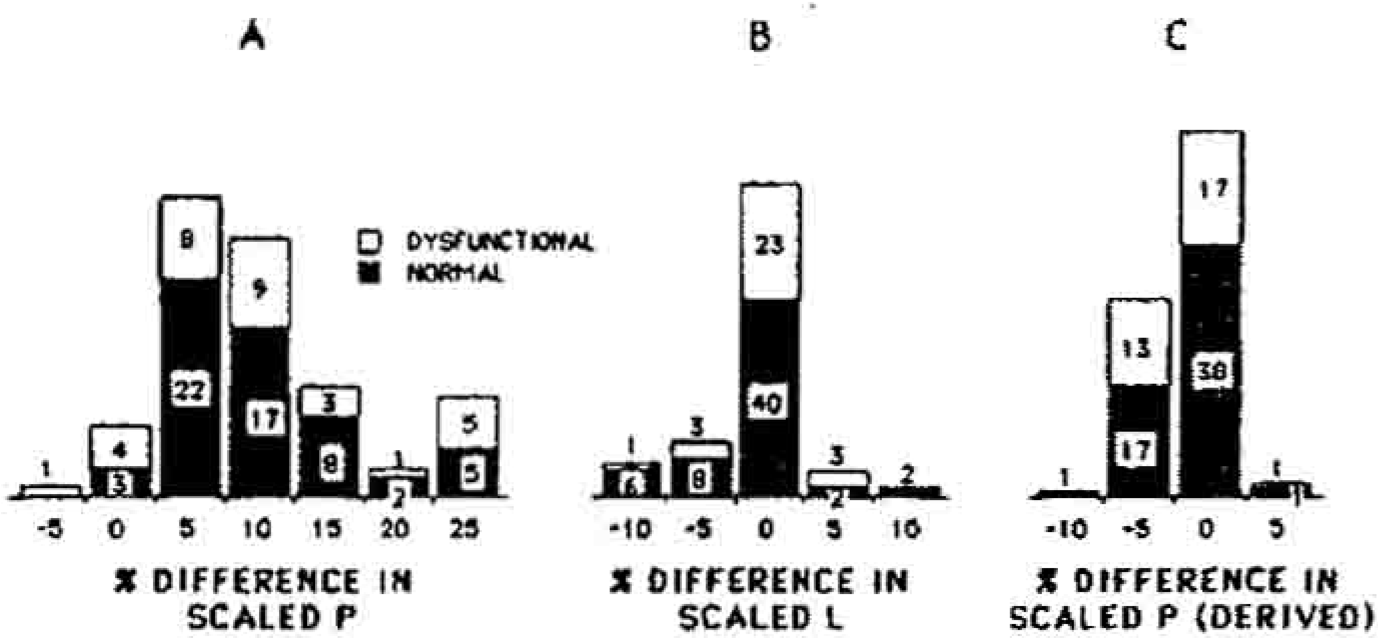



Fig. 2. Histograms showing the distribution of the percent difference between the au matic and manually filtered airflow waveforms. In all cases, the percent difference v scaled as a percentage of the manually filtered peak airflow value.

## Results. first experiment

We now consider the accuracy of the test system, using the manual inverse filter result as a standard. We collapse our results across loudness, sex and age in the following discussions, since scatter plots for the measures discussed indicated that accuracy did not vary significantly wilh any of these variables, except for a slight tendency toward more variability in the case of loud phonation.

Measurements of $T_o$ in almost all cases showed differences of less than two percent from measurements made from the EGG waveform. This degree of accuracy would be expected from the results reportcd previously for airflow-derived $T_o$ measurements (Rothenberg, 1977). As would also be expected measurements of mean airflow (M) made from the channel 2 signal were essentially the same as those from the manually inverse filtered signal, since the filtering procedures have no effect on the mean airflow. $Q_o$ measurements roughly agreed with the predictions from the EGG signal. but no quanlitative estimate of the correlation was derived, since the accuracy of the flow-derived $Q_o$ would depend greatly on the accuracy of the estimates of $L$ and $P$.

Thus, the parameters of most interest in these tests were the peak and minimum values of airflow. In Figure 2A, the value of peak airflow $P$ derived by the automatic procedure (channel 3) is compared with the value obtained by manual inverse filtering. The percentage error was computed assuming the manual procedure to be the reference or accurate condition. It can be seen that the errors were generally positive, resulting in values about 10% too high. This error occurred for both normal and disordered voices.

As shown in Figure 2B, the error for the minimum value $L$ was generally less than 5%, with the automated procedurc tending to give values slightly less than the manual filtering. As in Figure 2A, the percentage calculation was made with reference to the peak value of the manually inverse-filtered waveform, since this reference reflects the scale of interest for a particular waveform. (Since minimum values can be very small, or even zero, using the more accurate minimum value for the denominator would result in "error" percentages with little meaning.) As with the peak values in Figure 2A, the accuracy was generally maintained for both normal and disordered voices.

# Interpretation - first experiment

The error in minimum value, about 5% of peak flow, with a maximum of about 10%, would generally be considered adequate for clinical purposes. Variations of much more than 10% can be found among normal voices of the same sex and age and in a single voice within a sentence or at diffcrent times (Holmberg, el al., 1988; Karlsson. 1988; Schutte, 1980). The tendency for this error to be negative indicates that the errors may be largely due to remanent first formant energy not removed by the automated filtering. This might explain why there are proponionally more normal voices (with stronger F1 energy) that show the higher errors. Thus, some increase in the strength of the low-pass filtering, or a small reduction in the cutoff value, could conceivably reduce the error further and remove the negative bias.

The error in peak value P was of somewhat more concern, though the errors shown might still be acccptable for most applications. Because the error tended to be positive (from the approximately filtered waveform exceeding the assumed true glottal waveform), it was also deemed to be caused by some remanent F1 energy. This was verified by the data in Figure 2C, in which the peak of the more highly filtered Channel 2 waveform was used as the test value. It can be seen that the Channel 2 peak was generally within about 5 to10% of the accurately filtered value, with a slight tendency toward a too negative value, as would be caused by the overfiltering of the waveform. Thus, an optimum filter for peak value would lie somewhere between those used for Channel 2 and Channel 3. This hypothesis was supported by the additional experiment to be described below.

Thus, Figure 2 indicates that if the average error could be removed, an accuracy of 10% when compared to actual peak airflow can be attained by the new automated system in almost all cases, with most measurments within 5%. However, the presence of a few outliers with possible errors of over 15% was disturbing, since a presumed advantage of the new system was its robust procedure, that is, the absence of any feature that could cause a large error in unusual cases. To probe this potential problem further, a few of the outlying measurements were examined by comparing the print-outs of the waveforms in each channel of the DATA 6000. In cach case, the "error" was associated with a potentially incorrect manually inverse-filtered waveform; the vocalization did not have the long. clearly defined closed phase near zero flow that makes the inverse filter settings unambiguous. For example, in some cases a detailed examination of the waveforms suggestcd that the Channel 2 low-pass filtered waveform better preserved the true minimum glottal flow than did the presumably accurate, manually filtered waveform.

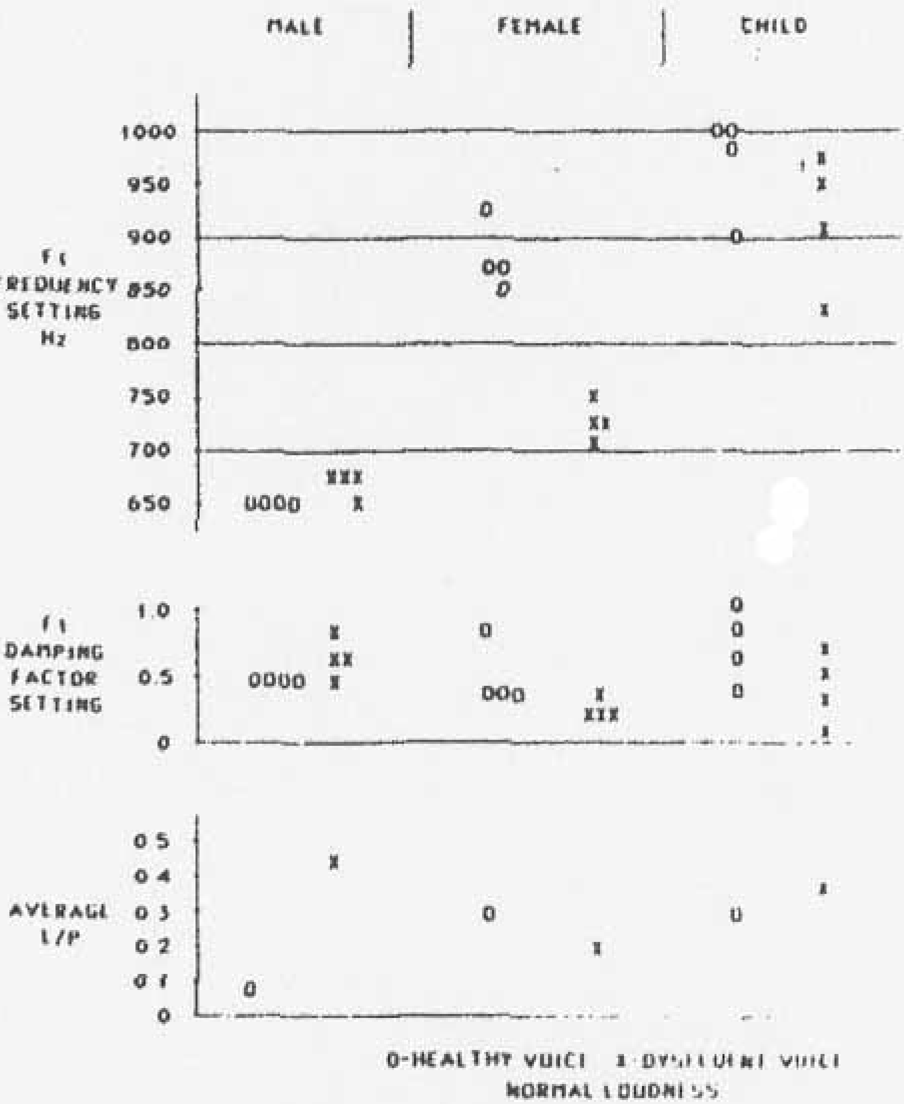



Fig. 3. Variation in the formant settings over four experimenters

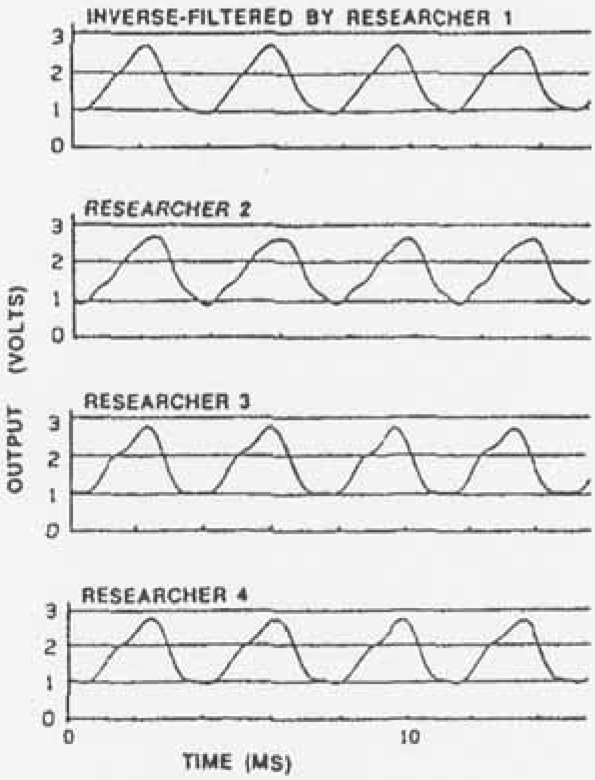



Fig. 4. Variation between adjusters in the manually adjusted inverse-filter output for the 11-year-old boy with a vocal fold nodule.

It therefore appeared to us that some significant proportion of the variance in the "errors" reported in Figure 2 was, in actuality, caused by errors in the parameters of the reference waveform. To investigate this possibility, as well as to test a revised filtering procedure in Channcl 3 for measuring *P*, as suggcstcd above, the following additional expcriment was performed.

# Method - reevaluation experiment

In this second, reevaluation experiment, data from six of the original subjects, chosen to represent the widest variety of glottal waveform types, were reprocessed with the system revised as shown by the dashed lines in Figure 1. The same analysis procedure was used, except that the manual inverse filtering for each sample was performed independently by four members of the research staff. including the two persons performing the previous inverse filter adjustments. Each adjuster had extensive experience in this task.

In the reviscd system, the filtering for Channel 3 was altered to include the approximate F1 inverse filter, and had a reduced low-pass setting, according to our interpretation of the results in Figure 2, A and B, above. In addition, the damping factor of the approximate Fl inverse-filter was changed from zero to 0.5 to match the approximate average vocal tract damping with the mask in place. The multiple versions of the manual inverse filtering were meant to give some indication of the variability possible in the manually set antiformants and the resulting variability in the reference values of *P* and *L*.

## Results - reevaluation experiment.

Results from the second experiment indicated that the biases in the estimation of both *P* and *L* are essentially removed in thc revised system. An increased variability in the error values was found. since some of the more difficult-to-inverse-filter voices were included in the sample of six subjects; however, an appreciable part of this variability appeared to be due to inaccuracy in the manual inverse filtering of the reference waveforms, as discussed above. This conclusion is supported by the fact that thc highest error values generally occurred with disordered voices that tended to be breathy. These waveforms usually had no clear, flat "closed" period near zero now in the inverse-filtered waveform to act as a reference in the adjustment procedure. In addition, informal observations with other subjects confirmed that little variance between experimenters is present when there is a clear closed phase with little or no airflow, as was the case for our sample of a healthy male voice.

The variability in the formant settings for breathy voices is shown in Figure 3. The first-formant settings (the most significant formant in determining the waveshape) are shown for all four experimenters for each of the six subjccts. Also shown as a measure of relative breathiness is the ratio *L*/*P*, as averaged over all reference values. This ratio will be zero if a complete glotta1 closure is attained during the closed period and approaches unity for very breathy voices. It appears from the figure that the variability in the formant settings is to some extent correlated with this measure of breathiness.

To show the effect on the waveform of the range of formant settings obtained by the different adjusters. Figure 4 presents the manually inverse-filtered waveforms from a vocalization by an 11-year-old boy diagnosed as having a vocal nodule. Though the resulting waveforms are grossly the same, there would be a significant variance in the resulting values for the minimum value *L* and, to a lesser extent, for the peak value *P*. It should be emphasized that without further knowledge there is no way to choose with confidence the most accurate waveform among the four. Even a waveform that shows some residual F1 energy near its minimum value could be correct, since there could be (and probably is) some F1 energy passing through the open glottis during that time interval.

## Reconstructing Idealized Waveforms

The airflow-based analysis system we envision would print out for each subject, in addition to the measured numerical parameter values, an idealized glottal airflow waveform that conforms to these values. This type of graphical printout would greatly simplify judgments of vocal function by making visually transparent the interrelationship of the various parameters and would also facilitate intra- and intersubject comparisons. In addition, when the analysis is performed separately for a number of consecutive glottal cycles, the resulting reconstructed waveform would exhibit more clearly the nature of any gross aperiodocities.

To test the viability of this type of graphical printout, the analysis results from three of the subjects were transferred manually from thc DATA 6000 system to a microcomputer which generated the required idealized waveform, given the measured parameter values. To conform to the truncated sinusoidal approximation of the glottal pulse described above, the idealized glottal volume-velocity *Ug* is defined by

$$Ug = [P - L / (1 - \cos(\pi Q_o)] * \cos(2\pi t)/(T_o) + P - [P - L/(1 - \cos(\pi Q_o)]$$

during the "open" periods, and remains at $L$ during the "closed" periods. This equation results in a symmetrical waveform that has the required values of $T_o$, $M$, $L$, $P$, and $Q_o$. To show a diversity of waveform types, the subjects chosen for this exercise were an adult male known to have a strong, efficient voice, the adult male Parkinson's disease patient, and the 7-year-old healthy female child.

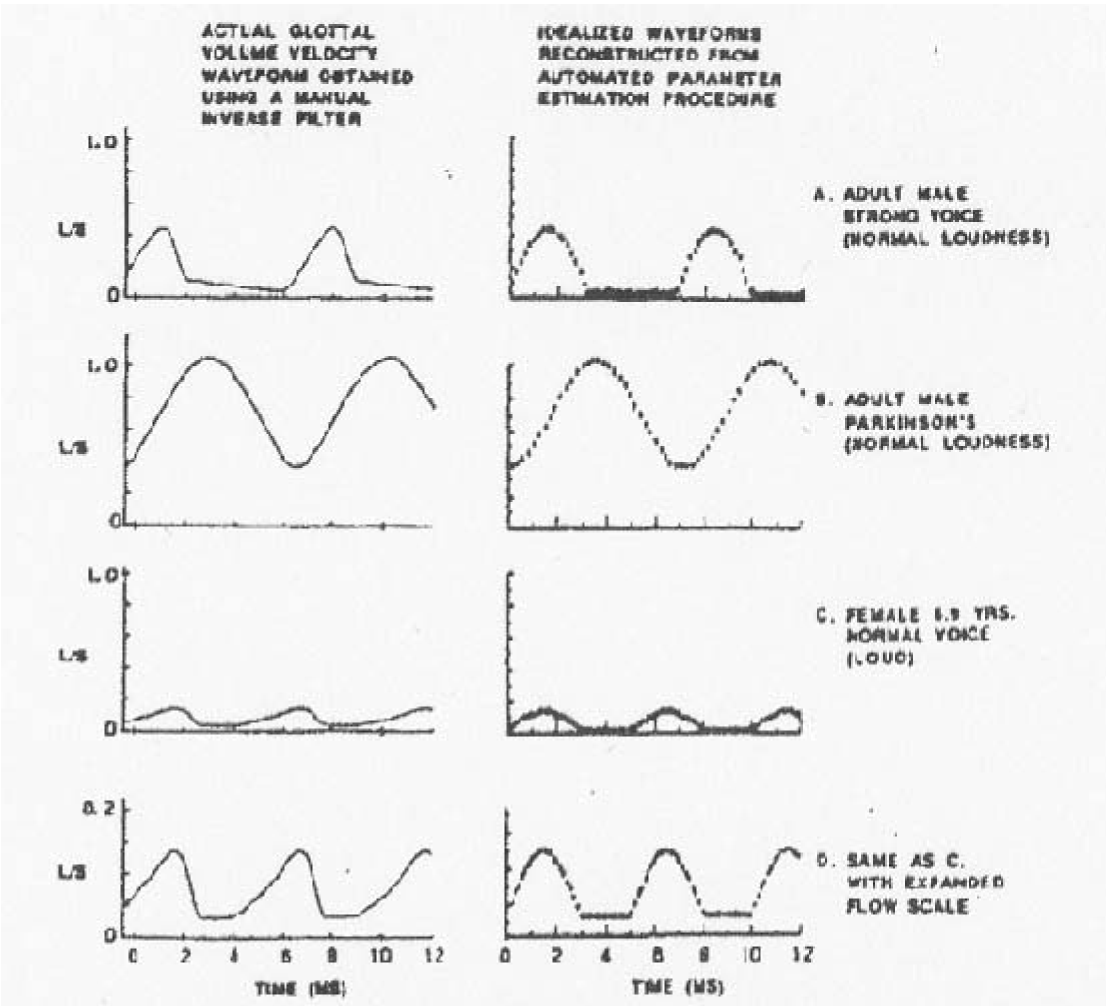



Fig. 5. Three repr sentative reconstruct glottal flow wavefor as compared with t output of the manua adjusted inverse-filter.

Parts A, B and C of Figure 5 compare the reconstructed glottal flow waveforms with the output of the manually-adjusted inverse-filter. The child's waveform is also shown with an enlarged flow scale, because of the much lower flow values. It can be seen from the figure that the reconstructed waveforms retain most of the significant properties of the manually obtained inverse-filtered waveforms, while eliminating many of the details such as a slight closed-period slope or remanent F1 energy - which would be of minimal interest to the clinician. The most notable exception is the asymmetry, or skewing to the right, of the glottal pulse that occurs in stronger voices; this is caused primarily by source-tract acoustic interaction and does not directly reflect vocal fold movements. However, if it is eventually found to be of interest clinically, this asymmetry could be inserted into the idealized waveform and the corresponding computation of $Q_o$, using a simple model of a source-tract interaction such as the one wruch was originally proposed by the author (Rothenberg, 1981) or a similar model proposed by Fant (1983). A measure of spectral balance or spectral slope for the mask waveform that reflected the relative strength of the higher frequency harmonics could also be used to help determine the degree of asymmetry, since a strengthening of the higher frequency harmonics is a primary correlate of this asymmetry.

# CONCLUSIONS

The results descrired above indicate that an automated parameter extraction system, similar to our revised system can re constructed that will have a standard deviation that is no more than 5% of the peak airflow value for measurements of both the minimum and peak flow values. This system will be robust in that it will rarely result in errors of more than about 10% of the peak flow value when used according to the designated protocol (mask, seal adjusted for no leakage, vowel similar to /æ/ or /a/). More precise determinations of system error than those made in this project will be difficult to obtain without some independent verification of the actual glottal airflow waveform that is more accurate than manual inverse-filtering by a highly trained operator. There is no method now available for such a verification in the human vocal tract, though a model experiment (mechanical, animal or computer) might be possible.

However, as pointed out above, a variance of 5%, or even 10%, in the measured values is not unreasonable for a clinical system. given the larger variance found among normal voices or in the same voice at different times. The other side of this coin must be that there are clinically significant variations in these airflow parameters that exceed 5% to 10%. This is generally acknowledged for average airflow, which has long been easy to measure, and evidence that this is also true for the parameters of peak and minimum airflow is evolving in current studies of breathy, hyperfunctional and aging voice (Fritzell, et al., 1983; Hillman. et al., 1988; Higgins, 1989).

The extrapolation of $Q_o$ from $M$, $L$ and $P$ also appears to be a reasonable alternative to other presently-proposed noninvasive procedures for estimating this variable.

# ACKNOWLEDGEMENTS

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